

Knee Extensor Capacity Utilization and Stair-gait Biomechanics in Younger and Older Adults

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners. I understand that my thesis may be made electronically available to the public.

Abstract

Stair-gait is a more challenging task than level-gait, particularly for older adults. Both ascending and descending stairs require greater lower extremity joint ranges of motion, vertical ground reaction forces, and external knee flexion moments than level-gait. Safe and efficient performance while negotiating stairs is largely dependent on adequate knee extensor muscle function. However, age-related declines to muscle quantity, quality, and function contribute to limitations in physical function and independence during activities of daily living, such as stair-gait. Consequently, compensatory movement strategies employed by the trunk may be a key mechanism by which those with reduced muscle function adapt to large knee joint demands encountered in daily activity. The large demands imposed by stairs paired with reduced muscle function among a growing ageing population poses a substantial threat to quality of life. A few studies have investigated knee extensor muscle utilization, a concept that describes the proportion of the maximal available capacity of the knee extensors that is used to complete a task. Knee extensor muscle utilization is greater in older adults compared to younger adults during walking and other tasks. A number of studies also demonstrate knee extensor muscle capacity utilization is larger during stair ascent and descent compared to level walking in both young and older adults. However, little data compare knee extensor muscle capacity utilization between young and older adults on stair negotiation; and no study has determined whether this is related to sagittal plane trunk kinematics.

This study aimed to compare knee extensor capacity utilization during stair ascent and descent in older adults versus younger adults. A secondary aim was to determine if this knee extensor capacity utilization is related to sagittal plane trunk kinematics. Twenty-five healthy young adults ($n = 16$) aged 18-30 years and healthy older adults ($n = 9$) aged 45-70 years were recruited primarily from the University of Waterloo and surrounding communities. Three-dimensional kinematics and kinetics were captured while participants ascended and descended a custom-built 4-step staircase using a high-speed motion capture system as well as floor- and step-embedded force plates, respectively. Knee extensor capacity utilization was calculated from maximal voluntary isometric and isokinetic contractions via

dynamometry and peak external knee flexion moments measured in the stance phase of stair-gait for both stair ascent and descent. A two-tailed Welch's t-test was used to determine if there was any difference in knee extensor capacity utilization between the healthy young adults and healthy older adults. A Pearson-Product Moment Correlation test was used to determine the relationship between knee extensor capacity utilization and the peak trunk flexion angle during the stance phase of stair-gait. Healthy older adults exhibited significantly greater knee extensor capacity utilization compared to healthy younger adults ($p = 0.02$), but not while descending stairs ($p = 0.62$). Knee extensor capacity utilization was not associated with peak sagittal trunk flexion angles during the stance phase of stair ascent ($r = -0.06$; $p = 0.77$) or stair descent ($r = -0.31$; $p = 0.14$).

Findings demonstrate that healthy older adults utilize a greater amount of their available knee extensor capacity than healthy younger adults during stair ascent but not during stair descent. KECU was not associated with peak trunk flexion while either ascending or descending the stairs. Overall, findings further support for stair ascent being more challenging for older adults than younger adults. This study provides insight into the impact of knee extensor weakness while negotiating stairs.

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List of Abbreviations

ABC-16	Activity-specific Balance Confidence Scale
ADLs	Activities of Daily Living
AMI	Arthrogenic Muscle Inhibition
BMI	Body Mass Index
BW	Body Weight
COM	Centre-of-Mass
COP	Centre-of-Pressure
CL & CTL	Controlled Lowering
GAQ	Get Active Questionnaire
EKFM	External Knee Flexion Moment
FC & FWC	Forward Continuance
FC & FCL	Foot Clearance
FP	Foot Placement
FPP	Foot Placement Preparation
HOA	Healthy Old Adult
HYA	Healthy Young Adult
IC	Initial Contact
KECU	Knee Extensor Capacity Utilization
KOOS	Knee Osteoarthritis and Outcome Scale
LP	Leg Pull-through
LS & LSW	Late-swing
MST	Mid-stance
MVIC	Maximal Voluntary Isometric Contraction
NPRS	Numeric Pain Rating scale
OA	Osteoarthritis
PD-Q	painDETECT Questionnaire
PU	Pull-up
SLS	Single-leg Stance
TO	Toe-off

vGRF	Vertical Ground Reaction Force
WA	Weight Acceptance
6MWT	6-Minute Walk Test
9SCT	9-Step Stair Climb Test

1 Introduction

Ageing is the process of declining physiological function, where the rate of entropy is greater than synthesis (Gilbert, 2000). Loss of skeletal muscle mass, known as sarcopenia, as well as the loss of skeletal muscle function, known as dynapenia, are two physiological aspects commonly related to the ageing process (Clark & Manini, 2012; Manini & Clark, 2011; Sayer et al., 2024). Age-related muscle atrophy, or sarcopenia, is often accompanied by dynapenia, though these terms have been thought to be conceptually distinct (Clark & Manini, 2008; Ohinata et al., 2024). Dynapenia can be described in several ways, including the decline in strength and, more recently, power (Clark & Manini, 2012; Freitas et al., 2024; Manini & Clark, 2011). In this context, strength is defined as the peak force or torque producing ability of a muscle, while power is defined as the maximum rate of work done by a muscle (Freitas et al., 2024; Knudson, 2009; Sayers, 2007). Both age-related sarcopenia and dynapenia contribute to declining physical function, independence, and quality of life (Manini & Clark, 2012; Ohinata et al., 2024).

Dynapenia impedes physical function during activities of daily living (ADLs) (Sayers, 2007). Safe and efficient performance of ADLs, including stair ambulation, is heavily reliant on knee extensor muscle function. Both stair ascent and descent are complex tasks requiring the knee extensors to help move, support, and stabilize the lower extremity (Bennell et al., 2008, 2013; Segal et al., 2009; Segal & Glass, 2011; Slemenda et al., 1998). With age, knee extensor muscle mass decreases more than knee flexors (Frontera et al., 2000; Frontera, Zayas, & Rodriguez, 2012). While knee extensor strength decreases with age, knee extensor power declines more rapidly and sooner with age than strength, and to a greater extent concentrically than eccentrically (Frontera et al., 2000; Hortobágyi et al., 1995; Sayers, 2007). A functional limitation such as challenge with stair-climbing limits participation in regular physical activity, likely forming a cycle of disuse and functional decline leading to increased multimorbidity risk.

Because they are weaker, older adults may need to utilize a greater proportion of their available knee extensor muscular capacity to meet the demands of ADLs (Hafer & Boyer, 2020; Hortobágyi et al., 2003).

Using a greater proportion of muscle capacity (termed knee extensor capacity utilization) could be further increased for those less physically active. Few studies, however, have attempted to understand how knee extensors are utilized during stair ascent and descent by both younger and older adults. Exploring how much of the total capacity of the knee extensor muscles during stair ascent and descent would expand on the existing literature on muscle function and ADLs. Furthermore, no study has assessed how the body compensates in situations where an individual is working near the maximum knee extensor capacity. It is possible that altering movement strategies to offload the knee extensors, such as positioning the centre of gravity over the knee joint to reduce the demand on the knee extensors, are implemented in the presence of knee extensor weakness. No study to date has examined whether knee extensor capacity utilization during stair ascent or descent is associated with sagittal plane trunk kinematics.

The purpose of this study is to investigate knee extensor capacity utilization and sagittal plane trunk kinematics while ascending and descending stairs in older and younger adults. The findings expand on current knowledge of muscle function and ageing while also providing insight into sagittal plane trunk strategies used to negotiate stairs and how it may link to overall muscle function. This data provides clinicians with a greater understanding of age-related declines to muscular function and serve as a foundation to identify potential means to help monitor and mitigate musculoskeletal disease initiation or progression.

2 Literature Review

2.1 Societal and Economic Burden of Aging

As the global populace ages, the prevalence of disease and chronic illnesses is likely to follow. In Canada, the population of older adults (65+ years) in 2019 was nearly 20% (6.6 million) of the overall population; and is estimated to increase to nearly 25% (10.7 million) of the population by 2040 (Public Health Agency of Canada, 2020). By 2070, the proportion of older adults and Canada is projected to be up to 32.3% (15.3 million) (Statistics Canada, 2024). With the ageing population increasing, the average age is also projected to increase from 40.1 years in 2023 to up to 50.1 years in 2073 (Statistics Canada, 2024). While the average age is increasing, quality of life may not follow the same trajectory.

Approximately 37% of older adults are estimated to report chronic disease comorbidity from the 2017–2018 Canadian Community Health Survey (CCHS) (Public Health Agency of Canada, 2020). The Canadian Chronic Disease Surveillance System estimates that multimorbidity, the presence of 2 or more chronic diseases, in adults 40 years and over increased from 20.5% in 2001 to 26.5% in 2011 (Feely, Lix & Reimer, 2017). Another study based on the Canadian Longitudinal Study on Aging found multimorbidity in adults aged 45-85 years to be 39.6% (Sakib et al., 2019). Furthermore, this study found multimorbidity to increase linearly with age and to be higher in women than men (Sakib et al., 2019). A systematic review estimated the prevalence of multimorbidity in those with dynapenia to be between 16-25.9% (Ohinata et al., 2024). Other studies have further demonstrated increased mortality and incident disability in those with lower knee extensor strength (Mitchell et al., 2012). This may suggest that a concurrently increasing ageing populace with a physical functional decline will likely result in a substantial societal and economic burden stemming from reduced quality of life as well as increased disability.

The process of ageing likely predisposes individuals to be more susceptible to musculoskeletal diseases such as osteoarthritis (OA) (Anderson & Loeser, 2010). Of those chronic diseases reported in the

2017-2018 CCHS, OA is the second highest ranking prevalence with 37.9% (Public Health Agency of Canada, 2020). OA is a prominent cause of disability associated with ageing in both males and females, with the hip, knee, and hand most commonly affected (Anderson & Felson, 1988; Arden & Nevitt, 2006; Felson et al., 1987). Age-related musculoskeletal decline and associated OA are accompanied by a substantial societal and economic burden that will increase with the growing ageing population (Bombardier et al., 2011; Sharif et al., 2015). For example, a population-based microsimulation model (POHEM-OA) estimated the total direct cost of OA in Canada will rise from \$2.9 billion in 2010 to \$7.6 billion in 2031 (Sharif et al., 2015). This study used a linear regression analysis to estimate total joint replacement costs from Canadian Joint Replacement Registry (CJRR) data between 2001-2011 (Sharif et al., 2015). According to annual statistics from the 2018-2019 CJRR, approximately \$1.4 billion was spent on hip and knee replacement surgeries alone (CJRR Annual Statistics Summary 2018-19, 2020). The POHEM-OA estimates hip/knee total joint replacement costs to increase to \$2.8 billion by 2031 (Sharif et al., 2015). These economic projections emphasize the importance of aiming to mitigate factors that contribute to disability, including physiological factors such as muscle weakness.

2.2 Skeletal Muscle Mass and Function

Numerous age-related changes to the myofibre of muscles are associated with the resulting loss of skeletal muscle mass (Sayer et al., 2024). These are processes heavily triggered or influenced by oxidative stress, impacting appropriate conditions for proteostasis ultimately leading to muscle mass loss (Sayer et al., 2024). Other physiological changes external to the myofibre may also influence muscle quality such as increased adipose tissue accumulation and alterations to anabolic hormonal regulation contributing to protein synthesis (Frontera, Zayas, & Rodriguez, 2012).

Quantifying muscle function is often achieved through the measurement of muscle strength and power. Together these two measures characterize muscle capacity; that is, the ability of a muscle to perform to the demands of a physical task (Brady & Straight, 2014). Normal muscle function is important

for mediating internal joint loads relative to external loads of a given task (Cole et al., 1996). Concerning the knee extensors, skeletal muscle functions to decelerate and accelerate the lower limbs, absorb loading forces, and provide dynamic joint stabilization (Bennell et al., 2008, 2013; Segal, Yack, et al., 2009; Segal & Glass, 2011; Slemenda et al., 1998). Lower muscle power is associated with functional limitations during ADLs to a greater extent than muscle strength (Skelton et al., 1994). The age-related decline of both muscle strength and power is often termed “dynapenia”, although these two do not appear to decline proportionately with age (Clark & Manini, 2012; Pickering & Chapurlat, 2020; Sayers, 2007).

2.2.1 Muscle Strength

Muscle strength is defined as the peak magnitude of force or torque development, or the capacity of a muscle to produce a maximal force. It is a function of a cross-sectional area as well as the pennation angle of muscles and its ability to recruit and fire descending alpha motor neurons to muscle fibres (Bennell et al., 2008; Bruce et al., 1997; Reid et al., 2015). Strength can be measured isometrically, isokinetically, or dynamically. Isometric strength is often measured against an immovable object (Macaluso & De Vito, 2004). To measure isometric strength of the muscle group acting at a joint, maximal voluntary isometric contractions (MVIC) are often measured via dynamometry in a fixed joint position (Macaluso & De Vito, 2004). Isokinetic strength is also often measured via dynamometry at a fixed rate (Macaluso & De Vito, 2004). Lastly, dynamic measures of strength often involve moving a maximal weight; for example, a 1 repetition maximum of a squat (Macaluso & De Vito, 2004; Reid & Fielding, 2011).

Generally, adults grow weaker with age. Peak knee extensor torques measured isometrically on a dynamometer were greater in younger adults than older adults (218.60 ± 34.79 and 128.61 ± 13.53 Nm, respectively; $p < 0.01$) (Jenkins et al., 2014). Two longitudinal studies have found the rate of decline in knee extensor strength to be between 1.8-2.0% per year (Frontera et al., 2000; Hughes et al., 2001). However, muscle strength may not decrease at the same rate as muscle mass with age. Declines in the

force producing capability of muscle generally decline to a greater extent and sooner than the cross-sectional area of muscle (Bruce et al., 1997). Longitudinal studies demonstrate the rate of muscle strength to decrease at a rate of 3-4% per year and 2.5-3% per year while the rate of muscle mass loss to decrease at a rate of 0.80-0.98% per year and 0.64-0.70% per year for older men and women, respectively (Mitchell et al., 2012; Wilkinson, Piasecki, & Atherton, 2018). Similarly, in a 5-year study of nearly 1700 older adults, 16% and 13% reductions to knee extensor strength but only 5% and 3% reductions to thigh muscle mass were found in men and women, respectively (Frontera et al., 2000). Reductions to skeletal muscle mass only reflect a small amount of the variance in changes to muscle strength (Clark & Manini, 2008). Conversely, increases in skeletal muscle mass only has a small influence on changes to muscle strength (Clark & Manini, 2008). This relationship is postulated to be stronger in younger adults, though with conflicting evidence (Macaluso & De Vito, 2004). This suggests that while sarcopenia influences muscular strength decline, it is also impacted by other age-related physiological changes.

2.2.2 Muscle Power

Muscle power is the rate of muscular work, or the ability to produce force rapidly (Accettura et al., 2015; Bean et al., 2002; Sayers, 2007). It is the product of dynamic force (or moment) production and muscle contractile velocity (Accettura et al., 2015; Bean et al., 2002; Reid & Fielding, 2012; Reid et al., 2015; Sayers, 2007). When measuring knee extensor power, the rate of moment development is given by the product of the knee extensor moment of force and the angular velocity of the knee joint (Calder et al., 2014). In this case, knee extensor muscle power is often measured using an isokinetic dynamometer at submaximal isotonic loads, although other methods have been used (Sayers, 2007). Dynamic power, similar to dynamic strength, can also be quantified as the force applied during a maximal weight test, such as a 1 repetition maximum of a squat, as long as the velocity of movement is also measured (Macaluso & De Vito, 2004). Vertical jumps on force platforms have also been used to assess muscle power (Izquierdo

et al., 1999).

Age-related change to knee extensor muscle volume may not impact strength and power equally (Davison et al., 2017; Segal & Glass, 2011). Peak muscle power declines more rapidly and begins earlier with age than muscle strength (Sayers, 2007; Skelton et al., 1994). Because muscle power appears particularly sensitive to aging, muscle power may be regarded as a better predictor of mobility, physical function, and knee joint load regulation than muscle strength (Accettura et al., 2015; Bean et al., 2002; Berger et al., 2012; Calder et al., 2014; Skelton et al., 1994). In a cross-sectional study of community-dwelling mobility-limited older adults ($n = 45$), Bean et al. (2002) found leg power to be a more significant predictor of physical function than leg strength, especially with stair climbing performance. In another study, frail older adults who required assistance with ADLs had 42-54% less leg extensor power compared to older adults who did not require assistance (Bassey et al., 1992; Macaluso & De Vito, 2004). A study of women with radiographic and symptomatic knee OA found a positive association between knee extensor muscle volume and isotonic knee extensor power [20% MVIC: $\beta = 0.624$; 95% CI = 0.129, 1.120; $p < 0.05$; 40% MVIC: $\beta = 0.669$; 95% CI = 0.294, 1.045; $p < 0.05$], but not knee extensor strength ($\beta = 0.001$; 95% CI = -0.002, 0.003; $p = 0.615$) after controlling for mean peak activation (Davison et al., 2017). As muscle power is measured as the product of muscle force and contractile velocity, physiological changes contributing to the latter are likely influenced by age (Reid & Fielding, 2011). This is partly a product of the proportion of type I to type II muscle fibres as well as the excitability of muscle (Macaluso & De Vito, 2004). In animal models, type II muscles fibres produce four times the power output than type I muscle fibres (Macaluso & De Vito, 2004). Furthermore, reductions to the size and amount of type II muscle fibres accelerates beyond the age of 50 (Lexell, Taylor, & Sjöström, 1988). In a study of healthy middle-aged ($n = 24$), healthy older adults ($n = 25$), and mobility limited older adults ($n = 23$), those with mobility limitations demonstrated longer time from electromyography (EMG) onset of the quadriceps and hamstrings to movement onset and lower rates of EMG rise (Clark et al., 2011). The same group of mobility-limited older adults also demonstrated 40-58% lower power output on the leg press than healthy

middle-aged and older adults ($p < 0.001$) (Clark et al., 2011). These findings suggest that structural and neuromuscular changes to muscle contribute to reductions to power and physical functional impairments associated with ageing.

2.3 Decrements in Knee Extensor Strength and Power & OA

Knee extensor weakness is common in those with pain and physical function impairments and may precede the degenerative structural joint changes associated with the initiation and progression of knee OA because of the influence of muscle on knee joint loads and stabilization (Amin et al., 2009; Bennell et al., 2008, 2013; Lewek et al., 2004; Øiestad et al., 2015; Segal et al., 2009; Segal et al., 2010; Segal & Glass, 2011; Slemenda et al., 1997, 1998). Age-related sarcopenia is thought to be accelerated in those with knee OA (Davison et al., 2017; Sayers, 2007).

An animal model showed knee extensor muscle weakness might be an independent risk factor for knee joint degeneration and subsequent OA development (Herzog & Longino, 2007). However, the mechanisms by which knee extensor muscle weakness occurs in those with knee OA is not fully understood (Amin et al., 2009; Segal et al., 2010; Segal & Glass, 2011; Sharma et al., 2003; Slemenda et al., 1998). It could be linked to various factors like muscle mass, proprioception, motor unit recruitment, excitation-contraction coupling, or activation inhibition (Segal et al., 2012). Poor knee extensor strength is also hypothesized to alter the location and magnitude of mechanical loading which could potentially lead to articular cartilage damage and the subsequent progression of knee OA (Sharma et al., 2003; Slemenda et al., 1998). However, the local mechanical environment of the knee joint, characterized by joint laxity and alignment, could be a more important determinant of the joint response to loading by muscle forces (Sharma et al., 2003).

Surprisingly, Sharma et al. (2003) found that greater baseline knee extensor strength was associated with increased likelihood of radiographic tibiofemoral OA progression in knees with greater valgus-varus malalignment and high-laxity than those with lower baseline knee extensor strength. Contrary to these

findings, Amin et al. (2009) found greater quadriceps strength did not increase the risk of tibiofemoral cartilage loss, even in knees with more varus malalignment, and to be protective against lateral patellofemoral cartilage loss but only for those with increase knee joint laxity. The latter used magnetic resonance imaging (MRI) which enables a direct measurement of cartilage loss compared to radiographic imaging which indirectly assesses cartilage loss by measuring joint space narrowing (Amin et al., 2009). Impairments in muscle function in knee OA are likely a combination of neuromuscular deficits and muscle atrophy (Bennell et al., 2013).

A mechanism linked to neuromuscular deficits in knee OA is arthrogenic muscle inhibition (AMI), which is the central nervous system's failure to voluntarily activate a muscle fully due to reduced afferent and efferent stimulation (Lewek et al., 2004; O'Reilly et al., 1998; Rafsanjani et al., 2017; Slemenda et al., 1998). AMI can be measured similarly to central activation ratios by taking the ratio of MVIC force to MVIC force superimposed with electric stimuli (SMVIC); a lower ratio indicates greater involuntary motor unit recruitment and higher AMI (Kent-Braun & Le Blanc, 1996; Rafsanjani et al., 2017). Another method to measure AMI can be taken by the ratio of the difference between the MVIC and SMVIC to the SMVIC; a higher ratio indicates greater involuntary motor unit recruitment and higher AMI (Rafsanjani et al., 2017). Reasons why AMI may occur in those with knee OA could be attributed to joint pain, effusion or swelling, joint damage, decreased motivation, or fear of further joint injury or pain (Bennell et al., 2013; Lewek et al., 2004; Rafsanjani et al., 2017). In one study, common peroneal nerve stimulation during an isometric leg extensor task reduced the quadriceps arthrogenic inhibition ratio indicating knee OA patients may be able to overcome possible inhibitory influences via indirect excitatory nerve stimulation (Rafsanjani et al., 2017). Nonetheless, the true reason for AMI occurrence in knee OA is mostly unknown.

Knee extensor power was a better predictor of mobility and stair task performance than knee extensor strength in a cross-sectional study of community-dwelling older adults with clinical knee OA (Accettura et al., 2015). Knee extensor power at lower loads and higher angular velocities may be a

stronger predictor of self-reported function in knee OA than MVIC torque and power at higher loads (Berger et al., 2012). Peak leg extensor power at low resistances (40% 1 Repetition Maximum (RM)) and high resistances (70% 1RM) have also been independently associated with less knee pain (both $p < 0.05$) and better quality of life ($p = 0.04$ and $p = 0.003$, respectively) in a sample of adults with symptomatic knee OA (Reid et al., 2015). When looking at how muscle power relates to knee joint loading, there are controversial findings. A cross-sectional study of community-dwelling adults with clinical knee OA found knee extensor power to have a positive association with knee adduction moment (KAM) peak and impulse during stride in gait, whereas knee extensor strength did not (Calder et al., 2014). Peak KAM is often used to reflect medial joint loading associated with worsening knee OA and is often reported to be elevated in those with knee OA (Calder et al., 2014). The positive association suggests greater knee extensor power could be disadvantageous for knee OA initiation or progression (Calder et al., 2014). The evidence would appear to support knee extensor power as an important factor for improving mobility, physical function, pain, and quality of life, despite potentially compromising medial joint loading during stride in gait (Accettura et al., 2015; Bean et al., 2002; Berger et al., 2012; Calder et al., 2014; Reid et al., 2015).

2.4 Stair-gait Biomechanics & Ageing

Negotiating stairs is more demanding than level-walking in younger adults (Costigan, Deluzio, & Wyss, 2002; McFadyen & Winter, 1988; Nadeau, McFadyen, & Malouin, 2003). Specifically, ascending and descending stairs demands greater joint range-of-motion (except for hip range-of-motion in stair descent), forces, moments, and powers (Andriacchi et al., 1980; Costigan, Deluzio, & Wyss, 2002; McFadyen & Winter, 1988; Nadeau, McFadyen, & Malouin, 2003; Riener, Rabuffetti, and Frigo, 2002). These demands are likely to be influenced by step inclination, which varies in the literature, and anthropometrics, which are rarely reported. Additionally, anteroposterior boundaries of a person's base of support are constrained by tread depth and tread depths do not necessarily account for all foot sizes (Bosse et al., 2012; Mian et al., 2007). These factors introduce complexity in summarizing or comparing

findings across studies. Regardless, difficulty with stair negotiation is reported to be present with over 20% of adults above the age of 50 and 45% of adults over the age of 80 (Startzell et al., 2000). With increasing age, it is likely that the large demands posed by stair negotiation, which are static, may approach or exceed one's functional capacity, which declines with age. Defining stair ascent and descent as well as summarizing the temporospatial, kinematic, and kinetic variables used in the literature will set a foundation of what is currently known about how people in different age groups negotiate stairs.

2.4.1 *Cycle Definitions of Stair-gait*

2.4.1.1 *Ascent*

One full stair-gait cycle for stair ascent can be divided into the stance phase and the swing phase, both of which can be further divided into sub-phases. Stance-to-swing ratios have been reported to be approximately 64:36, which is comparable to that of level-gait (McFadyen & Winter, 1988). Although the literature often does not specify the use of sub-phases for analyses, these sub-phases are used to characterize unique biomechanical features of stair-gait with greater specificity.

Stance. *Weight acceptance (WA)* is defined as the duration in which the stance limb has made initial foot contact with the step and accepts the weight of the body (McFadyen & Winter, 1988). During stair climbing, the initial contact being made on the foot is typically on the middle to anterior aspect of the foot and shifts posteriorly towards the heel as weight is being accepted (as opposed to the traditional heel contact phase during level gait in which the first point of contact is often at the posterior aspect of the foot and shifts anteriorly with weight acceptance) (Levangie & Norkin, 2011; McFadyen & Winter, 1988; Riener, Rabuffetti, and Frigo, 2002). *Pull-up (PU)* (sometimes referred to as “Push-Up”) is a single-limb support phase in which the stance limb is preparing to accelerate the body onto the next step (Levangie & Norkin, 2011). This phase occurs from the beginning of single-leg support to contralateral mid-swing (McFadyen & Winter, 1988). Through primarily concentric muscle contractions, the stance leg accelerates the body's mass against gravity (Riener, Rabuffetti, and Frigo, 2002). *Forward continuance (FC)* is from

ipsilateral mid-stance and contralateral mid-swing to ipsilateral toe-off and contralateral weight acceptance to accelerate the body upward and forward onto the next step (Levangie & Norkin, 2011; McFadyen & Winter, 1988).

Swing. *Foot clearance (FCL)* occurs from toe-off to mid-swing and resembles the point at which the foot has cleared the height of the step (McFadyen & Winter, 1988). *Foot placement positioning (FP)* occurs from mid-swing to late swing (McFadyen & Winter, 1988).

2.4.1.2 Descent

One full stair-gait cycle for stair descent can be divided into the stance phase and the swing phase. Similarly to stair ascent, stance and swing phases can be further divided into subphases, although there is variation in the literature regarding how these sub-phases are defined.

Stance. *Weight acceptance (WA)* occurs on ipsilateral initial foot contact to ipsilateral mid-stance and contralateral toe-off (McFadyen & Winter, 1988). This phase is sometimes also referred to as the “double-leg support” phase. *Forward continuance (FC)* marks the beginning of the single-leg support phase spanning from ipsilateral mid-stance and contralateral toe-off to contralateral mid-swing (McFadyen & Winter, 1988). *Controlled lowering (CL)* marks the second half of the single-leg support phase spanning from ipsilateral mid-stance and contralateral mid-swing to ipsilateral toe-off and contralateral initial foot contact resembling the stance-limb decelerating the weight of the body (McFadyen & Winter, 1988). Both FC and CL are frequently reported together simply as the “single-leg support” phase. Further division of the single-leg support phase into two additional sub-phases allows for a more comprehensive analysis of the distinct biomechanical features that occur during single-leg support.

Swing *Leg pull-through (LP)* occurs from toe-off to mid-swing and resembles the point in which the swing leg accelerates forward towards the next step (McFadyen & Winter, 1988). *Foot placement preparation (FPP)* occurs from mid-swing to late swing (McFadyen & Winter, 1988). It is important to

note that the swing phase is often reported as being a phase independent of any sub-phases.

2.4.2 *Temporospatial Characteristics of Stair-gait*

2.4.2.1 *Ascent*

On ascent, studies have demonstrated that older adults show slower step cadences in stair ascent compared to young adults (94.76 ± 13.03 and 102.50 ± 8.86 steps/min, respectively; $p < 0.31$) (Novak et al., 2011), findings were consistent with earlier work by Livingston (1995). However, other studies suggest a similar step cadence in younger and older adults [$(92 \pm 10$ steps/min) (Reeves, Spanjaard, et al., 2008), $(98 \pm 13$ and 92 ± 10 steps/min, respectively) (Reeves et al., 2009)]. When observed, differences in step cadences can be attributed to longer cycle times by older adults ($p \leq 0.03$) (Livingston, 1995). Although, two studies with a mean age of approximately 28 years of age found cycle times to be between 1.41 and 1.45 seconds (Riener, Rabuffetti, and Frigo, 2002). This is in comparison to another study with a mean age of 53 years of age which found cycle times of 1.30 seconds (Nadeau, McFadyen, & Malouin, 2003). Faster cycle times by older adults could be attributed to differences in experimental set-up.

While the literature presents varied findings on cadence during stair-gait, it appears consistent that older adults spend greater time in the stance phase compared to younger adults (0.74 ± 0.08 and 0.82 ± 0.13 seconds, $p = 0.037$) (Novak & Brouwer, 2011). The stance phase requires considerable demands from the lower extremity extensor muscles to both accept and accelerate one's body mass. Slower stance phase times employed by older adults might indicate difficulty effectively negotiating stair ascent, leading to slower step cadences.

2.4.2.2 *Descent*

Slower step cadence in stair descent has been shown in older adults compared to young adults (103.68 ± 10.24 and 110.64 ± 10.24 steps/min, respectively; $p < 0.31$) (Novak et al., 2011). Lower step cadences by older adults were also reported in a previous study (94 ± 12 steps/min) (Reeves, Spanjaard et

al., 2008). One study reported longer cycle times in older adults compared to young adults (1.10 ± 0.16 and 0.96 ± 0.12 seconds, respectively; $p < 0.05$) (Mian et al., 2007). However, as in stair ascent, these cycle times demonstrated by older adults are shorter than those from a sample of 10 younger adults (1.19 ± 0.10 seconds) reported by Riener, Rabuffetti, and Frigo (2002). Slower step cadences could largely be attributed to differences in stance phase duration (Buckley et al., 2013; Novak & Brouwer, 2011; Livingston, 1995; Mian et al., 2007). Novak & Brouwer (2011) found stance times employed by older adults to be lower than younger adults (0.75 ± 0.14 and 0.70 ± 0.08 seconds, respectively; $p < 0.001$). Interestingly, another study found stance time variability to be greater in younger adults than older adults (Livingston, 1995). Less variability could indicate older adults descend stairs with a more constrained neuromuscular control strategy. Another study found that both weight acceptance (0.258 ± 0.079 and 0.226 ± 0.055 seconds, respectively; $p = 0.016$) and controlled lowering (0.624 ± 0.076 and 0.544 ± 0.071 , seconds, respectively; $p < 0.001$) phase times were significantly longer for older adults compared to younger adults (Buckley et al., 2013). However, one study did not find any differences between age groups for both single- and double-leg support, reflecting controlled lowering (combined with forward continuance) and weight acceptance, respectively (Bosse et al., 2012). Another study found only single-leg support to be significantly longer for older adults than younger adults (0.84 ± 0.12 and 0.72 ± 0.08 seconds, respectively; $p < 0.05$) (Mian et al., 2007). Conflicting findings may reflect differences in sample characteristics including physical activity level, lower extremity muscular capacity, or confidence and comfort with negotiating stairs, as well as differences in stair parameters such as tread width, depth, slope, and differences in protocols such as instructions provided to the participants.

2.4.3 *Kinetics & Kinematics*

2.4.3.1 *Ascent*

Kinematics. The kinematics of the lower extremity operate distinctly for stair ascent than descent since the direction of vertical movement is upwards instead of downwards. During the stance phase of

stair ascent, the knee starts in a position more flexed than level walking and extends to accelerate the body upward and forward onto the next step (Kaufman et al., 2001). The hip and ankle also move from a flexed and dorsiflexed position, respectfully, into an extended and plantarflexed position, respectfully (Propapadaki et al., 2007). Ankle, knee, and hip joint flexion angles increase with increasing inclination (Riener, Rabuffetti, and Frigo, 2002). Peak knee flexion and extension angles as well as peak ankle dorsiflexion and plantarflexion angles did not differ between older and younger adults during stair ascent ($p > 0.05$) (Reeves et al., 2009). However, older adults use less knee range of motion than younger adults during stance ($53^\circ \pm 6^\circ$ and $57^\circ \pm 3^\circ$, respectively; $p < 0.05$) (Hortobágyi et al., 2003). Reduced joint range-of-motion in old adults could reflect decreased joint mobility or co-contraction of antagonistic muscles to increase stability and control of movement.

Kinetics. Vertical ground reaction forces (vGRF) during ascent appear to have a double peak, akin to level gait albeit with a greater second peak, coinciding with weight acceptance and toe-off (Stacoff et al., 2005; Reeves et al., 2009). Reeves et al. (2009) found older adults demonstrated a lower first peak vGRF compared to younger adults (0.96 ± 0.04 and 1.03 ± 0.07 N/kg, respectively; $p < 0.01$). This was also demonstrated in a study sample of only women (1.04 ± 0.07 and 1.16 ± 0.08 BW; $p \leq 0.05$) (Hamel et al., 2005). These findings agree with another study, but for both initial contact (1.20 ± 0.11 and 1.27 ± 0.17 N/kg, respectively; $p < 0.05$) and push-off (1.79 ± 0.18 and 2.13 ± 0.23 N/kg, respectively; $p < 0.05$) the difference was reversed when forces were normalized to stair height (Bertuccio & Cesari, 2009).

Stair ascent requires concentric activation of the ankle plantar flexors, knee extensors, and hip extensors (McFadyen & Winter, 1988; Riener, Rabuffetti, and Frigo, 2002). Of the lower extremity, the knee extensors contribute the most to support stair ascent with external knee flexion moment (EKFM) reported to be 3 times greater than gait (Andriacchi et al., 1980; McFadyen & Winter, 1988). Contrary to vGRF, EKFM demonstrates a single peak coinciding with the initiation of the pull-up phase of stance (Hortobágyi et al., 2003; Novak & Bouwer, 2011; Reeves et al., 2009). Similarly to vGRFs, older adults have demonstrated a lower peak EKFM than younger adults (0.89 ± 0.22 and 1.19 ± 0.24 Nm/kg,

respectively; $p < 0.01$) (Reeves et al., 2009). This finding was also found in two other studies. Hortobágyi et al. (2003) found the EKFM employed by older adults to be 1.00 ± 0.22 Nm/kg compared to 1.55 ± 0.24 Nm/kg by younger adults ($p < 0.05$). A study from Novak and Bouwer (2011) found the EKFM employed by older adults to be 0.99 ± 0.21 Nm/kg compared to 1.02 ± 0.20 Nm/kg by younger adults ($p < 0.01$). These findings suggest that older adults may not use the knee extensors to effectively accept weight and pull the body up onto subsequent steps.

2.4.3.2 Descent

Kinematics. During stair descent, the knee starts in a nearly extended position and gradually flexes to lower the body (Kaufman et al., 2001). The hip and ankle also move into increased flexion and dorsiflexion, respectfully (Propapadaki et al., 2007). Reduced range-of-motion or ability to change joint angles quickly could lead to prolonged phase times. In single-leg support, older adults demonstrate significantly less knee flexion compared to younger adults ($20.3^\circ \pm 7.9^\circ$ and $29.7^\circ \pm 6.7^\circ$, mean $\pm$ SD, respectively) (Bosse et al., 2012). Furthermore, significantly higher peak vertical center-of-mass (COM) accelerations employed by younger adults compared to older adults (-2.20 ± 0.58 and -1.45 ± 0.42 m/s², mean $\pm$ SD, respectively) (Buckley et al., 2013). Similar findings were reported in a more recent study (Dixon et al., 2018). Older adults have slower center-of-pressure (COP) velocity when descending stairs compared to younger adults ($p < 0.01$) (Kim, 2009). Mian et al. (2007) found no differences in COM velocity between younger and older adults, although the physical activity level and confidence of the older adult sample was cited as being high compared to the average older adult.

Kinetics. Contrarily to level-gait and stair ascent, vGRFs have been reported to inconsistently demonstrate a second peak occurring in controlled lowering of stair descent indicating variations in loading strategy during this phase (Stacoff et al., 2005). Studies reporting double-peaks found greater first peak vGRF but lower second peak vGRF in stair descent compared to both stair ascent and level gait (Hamel et al., 2005; Riener, Rabuffetti, and Frigo, 2002). A sample consisting of only women found

second peak vGRF to be greater in older women compared to younger women (0.95 ± 0.07 and 0.89 ± 0.08 BW, respectively; $p \leq 0.05$) (Hamel et al., 2005).

Contrary to stair ascent, stair descent requires eccentric activation of ankle plantar flexors, knee extensors, and hip extensors. Eccentric contractions of the ankle plantar flexors are largely responsible for absorbing loads at initial contact during stair descent (Buckley et al., 2013). Older adults have been cited as possessing a reduced capacity to eccentrically load the plantar flexors at high angular velocities at initial contact (Buckley et al., 2013). Similarly to stair ascent, an initial peak EKFM occurs in 0-30% stance during weight acceptance, but a second peak EKFM has been reported to occur in 30-60% stance in stair descent only (Hunt et al., 2024; Kaufman et al., 2001; Lin et al., 2015; McFadyen & Winter, 1980; Riener, Rabuffetti, and Frigo, 2002). This second peak can be explained by the trailing knee flexing and shifting anteriorly beyond the center of pressure during controlled lowering, causing an increased knee extensor moment (Riener, Rabuffetti, and Frigo, 2002). Descending stairs also require a greater EKFM than ascending stairs or level gait (Kaufman et al., 2001; Novak & Brouwer, 2011; Riener, Rabuffetti, and Frigo, 2002). Although, one study found EKFM to be lower in stair descent than stair ascent in both younger and older adults (Hortobágyi et al., 2003). Novak and Bouwer (2011) found peak EKFM during controlled lowering to be greater for older adults than younger adults (1.15 ± 0.18 and 1.03 ± 0.21 Nm/kg, respectively; $p = 0.02$). However, a previous study found older adults to demonstrate lower peak EKFM than younger adults (0.64 ± 0.24 and 0.90 ± 0.24 Nm/kg, respectively; $p < 0.05$) (Hortobágyi et al., 2003).

Several studies have also recorded co-activation of the quadriceps and hamstrings during stair descent via EMG (Hortobágyi et al., 2003; Chandran et al., 2019). Hortobágyi et al. (2003) found hamstring activity to be greater in older adults than younger adults, yielding co-activation ratios of (46 ± 37 and 64 ± 32 %, respectively, $p < 0.05$). Another study also found co-contraction to be 18% greater in older adults than younger adults ($p < 0.001$) (Chandran et al., 2019). These findings might indicate that greater knee stiffness, reduced rate of knee flexion, and greater co-contraction at the knee could be a functional mechanism used by older adults to support balance and stability while negotiating stairs, while

consequently increasing phase durations, joint stress, and trunk flexion at phase initiation (Bosse et al., 2012; Buckley et al., 2013; Hunt et al., 2024).

2.4.4 *Trunk Kinematics*

Peak sagittal trunk inclination during stair ascent is twice that during stair descent and six times that during level-gait (Krebs et al., 1992). Though few studies have assessed this mechanism in ageing populations, sagittal plane trunk flexion could be an important means of minimizing the required EKFM by shortening the moment arm and reducing the magnitude of the vertical GRF vector (Asay et al., 2009; Hinman et al., 2002; Iijima, Eguchi, et al., 2019; Iijima et al., 2018; Trinler et al., 2016). Increases to trunk flexion could also be a result of reduced knee range-of-motion as well as increased stiffness and co-activation (Bosse et al., 2012). In clinical populations, this might serve as a compensatory mechanism to mitigate reductions to knee extensor muscle strength and quality often seen in clinical populations (Asay et al., 2009). For example, a negative correlation between peak sagittal trunk flexion and peak EKFM has been seen in those with more severe knee OA during stair ascent ($R^2 = 0.590$; $p = 0.006$) (Asay et al., 2009). A meta-analysis also revealed those with knee OA to have increased sagittal trunk flexion angles ($^{\circ}$) compared to controls in stair ascent (pooled SMD = 0.38; 95% CI = 0.02, 0.75; $p = 0.040$) (Iijima et al., 2018). However, significant differences in peak sagittal trunk flexion angle ($^{\circ}$) were only found when comparing healthy controls to more severe knee OA (13.4 ± 4.6 vs. 19.7 ± 10.1 , respectively, mean $\pm$ standard deviation, $p < 0.05$) (Asay et al., 2009).

During stair descent, increased trunk flexion could potentially reduce one's ability to effectively resist anterior movement, causing greater anterior COM accelerations (Bosse et al., 2012). Greater trunk flexion angles by older adults ($11.1^{\circ} \pm 5.8^{\circ}$) compared to younger adults ($8.0^{\circ} \pm 4.3^{\circ}$) have been reported in stair descent (Bosse et al., 2012). Similarly, greater COM-COP, defined as the angle formed by the line connecting the COP and COM with a vertical line through the COP, sagittal anterior and posterior angles were found to be greater in older adults than younger adults for both stair ascent and descent (Lee &

Chou, 2007). Consequently, increased trunk flexion could bring the body's COM in front of the body's COP and possibly outside of the base-of-support (BOS). This may further lead to increased instability and fall risk, particularly in an older population. A cluster analysis has identified different strategic phenotypes used by older adults while descending stairs based on trunk, hip, knee, and ankle angles during controlled lowering (Tanaka et al., 2025). Three clusters were identified: cluster one identified those with a more neutral posture ($n = 20$), cluster two identified those with increased hip and trunk extension and decreased knee extension ($n = 43$), and cluster three identified those with greater stance-side rotation angles of the trunk and lower extremity (Tanaka et al., 2025). Cluster two demonstrated trunk extension angles of 4.9° compared to 0.45° and 1.8° by cluster one and three, respectively. Interestingly, those in clusters two and three scored lower in standing balance and muscle strength factors on the Community Balance and Mobility Scale than cluster one (Tanaka et al., 2025). However, no significant differences were found in hip extension, hip abduction, and knee extension MVIC (Tanaka et al., 2025). Despite this, those in cluster two were found to have a greater COM-COP separation distance than the other two clusters (Tanaka et al., 2025). In healthy adults, the COM-COP separation distance has been found to be greater in stair descent than ascent (Zachazewski et al., 1993). These findings might suggest that sagittal trunk kinematics, specifically trunk extension, may be a strategy utilized by older adults with poor balance or muscular control patterns to minimize anterior COM accelerations and better control the lowering phase of stair descent (Bosse et al., 2012; Hamel & Cavanagh, 2004; Lee & Chou, 2007; Tanaka et al., 2025).

2.5 Muscle Capacity Utilization

A novel biomechanical perspective that could aid in explaining the relationship between muscle function and physical function is muscle capacity utilization (also termed “relative effort” and “functional demand” in the literature). This term could be used as a method of quantifying how much of a muscle's capacity is being used to meet the demands of a task (Hafer & Boyer, 2020; Hortobágyi et al., 2003). It can also be thought of as a means of characterizing muscular mechanical reserve (Hafer & Boyer, 2020). In practice, muscle capacity utilization could be an indicator of muscle function, such that a very high

muscle capacity utilization (suggesting the task requires a very large proportion of the total capacity of the muscle) could be a potential indicator of risk of musculoskeletal disease onset or progression such as knee OA. In this context, those with greater muscle capacity utilization operate closer to their maximal capacity, indicating poorer physical function. A greater ability to take-on task-imposed knee joint demands will promote long-term physical activity participation, an important factor in delaying the onset of or slowing the progression of knee OA. However, this conceptualization has rarely been used in clinical biomechanics.

The earliest known conception of muscle capacity utilization, prior to any formal recognition in academic literature, came from a group investigating the chair-rise task (Hughes, Myers, & Schenkman, 1996). The investigators compared the maximal knee extensor moment in the chair-rise to the MVIC of the knee extensors measured at 60° of knee flexion by dynamometry (Hughes, Myers, & Schenkman, 1996). At chair heights matching knee height, the relative effort of the knee extensor was $34 \pm 6.7\%$ for younger adults ($n = 10$) and $78 \pm 23.2\%$ for older frail adults ($n = 11$) ($p < 0.0002$) (Hughes, Myers, & Schenkman, 1996). At lower chair heights, relative effort of the knee extensors increased to $39 \pm 8.1\%$ for younger adults and $97 \pm 22.8\%$ for older adults ($p < 0.0002$) (Hughes, Myers, & Schenkman, 1996). A study from Hortobágyi and colleagues (2003) similarly quantified, and coined, “relative effort” as a ratio of the peak knee joint moments observed during different ADLs (chair-rise, stair ascent, & stair descent) to the MVIC of the knee extensors; however, MVICs were measured at 15°, 30°, 45°, 60°, and 75° of knee flexion on a leg press machine (Hortobágyi et al., 2003). Since MVICs were measured at different knee flexion angles, the investigators took the MVIC at the knee flexion angle closest to the knee flexion angle at which the maximal knee extensor moment occurred during the ADL as the denominator to the relative effort ratio (Hortobágyi et al., 2003). The relative effort of the knee extensor for the chair-rise task was $42 \pm 19\%$ for younger adults ($n = 13$) and $80 \pm 34\%$ for older adults ($n = 14$) ($p < 0.05$) which is comparable to previous work (Hughes, Myers, & Schenkman, 1996; Hortobágyi et al., 2003). In stair ascent, relative effort of the knee extensors was $54 \pm 16\%$ in younger adults and $78 \pm 20\%$ in older adults ($p < 0.05$)

(Hortobágyi et al., 2003). In stair descent, relative effort of the knee extensors was $42 \pm 20\%$ for younger adults and $88 \pm 34\%$ for older adults ($p < 0.05$) (Hortobágyi et al., 2003). This data demonstrates that older adults operate at nearly twice the relative effort than younger adults while performing ADLs (Hortobágyi et al., 2003).

Subsequent studies began using maximal voluntary isokinetic contractions at various angular velocities as the maximal capacity of the knee extensors. Reeves et al. (2008; 2009) measured relative effort as the maximal knee extensor moment observed during stair ascent and descent with respect to the maximal isokinetic voluntary contractions of the knee extensors at various speeds. Knee joint angular velocities were measured by dynamometry at 60, 120, 180, and 240 °/s both concentrically (stair ascent) and eccentrically (stair descent) (Reeves et al., 2008; 2009). Comparable to methods from Hortobágyi et al. (2003), the investigators took the maximal isokinetic moment at the angular velocity that most closely corresponded to the angular velocity at which the maximal knee joint moment occurred during stair ascent or descent (Reeves et al., 2008; 2009). In stair ascent, relative effort of the knee extensors was $53 \pm 19\%$ for younger adults ($n = 17$) and $75 \pm 30\%$ for older adults ($n = 15$) ($p < 0.05$) (Reeves et al., 2009). In stair descent, relative effort of the knee extensors was $30 \pm 13\%$ for younger adults ($n = 17$) and $42 \pm 16\%$ for older adults ($n = 15$) ($p < 0.05$) (Reeves et al., 2008). Reeves et al. (2008; 2009) further measured plantarflexor relative effort and demonstrated that younger adults employ a relative effort of $85 \pm 33\%$ and $74 \pm 33\%$ while older adults employ $93 \pm 47\%$ and $77 \pm 25\%$, for both stair ascent and descent, respectively.

The term “functional demand” was introduced by Samuel and colleagues (2011), which expanded upon relative effort both conceptually and operationally. Functional demand was measured as the ratio of the peak joint moments observed during different ADLs to the estimated MVIC at the joint angle at which the peak joint moment occurred (Samuel et al., 2011). Knee joint flexion and extension MVICs were measured at 20°, 60°, and 90° of knee flexion on a custom-built dynamometer; hip joint flexion and extension MVICs were also measured at 0°, 30°, and 45° of hip flexion (Samuel et al., 2011). These

values were linearly interpolated to be able to reflect and estimate maximal muscle strength at any joint angle observed (Samuel et al., 2011). The investigators found the functional demand of the knee extensors was $103 \pm 39\%$ and $120 \pm 45\%$ for older adults ($n = 84$) in both stair ascent and descent, respectively (Samuel et al., 2011). The higher values reported by Samuel et al. (2011) than previously reported values of relative effort could largely be attributed to the method by which maximal muscle capacity was measured. Isometric muscle capacity does not accurately reflect muscle contractions with movement and may have contributed to underestimated maximal values, resulting in an overestimated functional demand metric (Samuel et al., 2011). Regardless, this research also reported on knee flexor ($42 \pm 16\%$ and $73 \pm 48\%$), hip extensor ($89 \pm 41\%$ and $51 \pm 35\%$), and hip flexor ($43 \pm 19\%$ and $43 \pm 43\%$) functional demands for both stair ascent and descent, respectively (Samuel et al., 2011). These findings suggest that older adults utilize a greater proportion of their available muscular capacity than younger adults when performing activities of daily living, including stair ascent and descent. Therefore, younger adults may be better equipped to both concentrically pull the body up onto the next step during stair ascent and eccentrically lower the body down to the next step during stair descent. Furthermore, relative effort or functional demand is relatively comparable for old adults across studies for stair ascent. However, there are controversial findings for stair descent due to differences in methods chosen to measure maximal muscle capacity which may not accurately reflect eccentric muscle action. This makes it difficult to ascertain whether stair ascent or descent requires a greater proportion of available muscle capacity.

Further building upon the concept of functional demand, Hafer & Boyer (2020) seemingly combine previously defined methods of relative effort and functional demand to formulate an approach which better reflects the dynamic muscle action observed during ADLs. Hafer and Boyer (2020) measured functional demand as the ratio of the peak joint moment observed during level-gait to the maximal joint moment at the joint angular velocity during which the peak joint moment occurred. Maximal voluntary knee extensor contraction was measured isometrically at 60° of knee flexion as well as isokinetically at $90^\circ/\text{s}$ and $270^\circ/\text{s}$ both eccentrically and concentrically by dynamometry (Hafer &

Boyer, 2020). These values were fitted with a third-order polynomial to estimate the maximal joint moment at any observed angular velocity (Hafer & Boyer, 2020). Although no study has used this method for either stair ascent or descent, using this method in level-gait has demonstrated that functional demand in less active older ($n = 19$) adults is greater than younger adults ($n = 20$) ($24.2 \pm 10.7\%$ and $20.4 \pm 10.2\%$, respectively, $p < 0.05$) and increases with faster walking speeds ($34.4 \pm 15.7\%$ and $28.9 \pm 12.4\%$, respectively, $p < 0.05$) (Hafer & Boyer, 2020). Thus, it was “more work” for the less active older adults to complete walking. Despite these results, the healthy older adults maintained a functional reserve of approximately 65% of knee extensor strength capacity at peak knee flexion moment during level walking (Hafer & Boyer, 2020).

It is unclear if baseline functional demand is a limiting factor in older adults with skeletal muscle disorders or in conditions where decreased muscular capacity is common, such as in knee OA. Other factors such as mobility and range of motion may also limit the ability of a muscle to be utilized effectively. The concept of muscle capacity utilization emphasizes the importance of considering muscular contributions during ADLs. Due to the limited available research and inconsistent measurement methods used, it is difficult to effectively summarize findings cohesively. However, one consistency in these findings is that older adults tend to use a greater proportion of their available knee extensor capacity and therefore are operating close to their functional limits. With age-related declines of muscle function, increases in demand may exceed functional limits further reducing effective ADL performance.

Table 2.1: Summary of the literature on both the measurement methods and estimations of muscle capacity utilization of the knee extensors during stair-gait.

Author	Sample	Muscle Capacity Measure	Muscle Capacity Estimation	Muscle Capacity Utilization
(Hortobágyi et al., 2003)	YA (n=13) OA (n=14)	Maximal voluntary isometric contractions at 15°, 30°, 45°, 60°, and 75° of knee flexion via leg press machine.	Used maximal capacity closest to the joint angle at which the maximal knee extensor moment occurred.	Ascent: YA = $54 \pm 16\%$ OA = $78 \pm 20\%$ Descent: YA = $42 \pm 20\%$ OA = $88 \pm 34\%$
(Reeves et al., 2008)	YA (n=17) OA (n=15)	Maximal voluntary isokinetic eccentric contractions at 60, 120, 180, and 240 °/s via dynamometry.	Used maximal capacity closest to the joint angular velocity at which the maximal knee extensor moment occurred.	Descent: YA = $30 \pm 13\%$ OA = $42 \pm 16\%$
(Reeves et al., 2009)	YA (n=17) OA (n=15)	Maximal voluntary isokinetic concentric contractions at 60, 120, 180, and 240 °/s via dynamometry.	Used maximal capacity closest to the joint angular velocity at which the maximal knee extensor moment occurred.	Ascent: YA = $53 \pm 19\%$ OA = $75 \pm 30\%$
(Samuel et al., 2011)	OA (n=84)	Maximal voluntary isometric contractions at 20°, 60°, and 90° of knee flexion via custom-built dynamometer.	Used linear interpolation to estimate the maximal capacity at the same joint angle at which the maximal knee extensor moment occurred.	Ascent: OA = $103 \pm 39\%$ Descent: OA = $120 \pm 45\%$

YA = Younger Adult; OA = Older Adult.

4 Literature Gaps

- 1 Knee extensor weakness is implicated in functional performance decline with ageing. Yet, there is little research using muscle capacity utilization (also termed “relative effort” or “functional demand”) as a biomechanical tool to assess knee extensor function and ADL performance. It is unknown if operating at a greater muscle capacity utilization may be associated with adaptive or compensatory strategies to accomplish ADLs. Furthermore, no study known has measured knee extensor capacity utilization during stair-gait using the moment-velocity relationship, as introduced by Hafer & Boyer (2020). Assessing knee extensor capacity utilization during stair-gait in both older and younger adults using this method would help quantify the underlying mechanisms linking muscle capacity deficits seen with ageing and performance of mobility tasks common in daily living.
- 2 There is limited research investigating sagittal plane biomechanics during stair-gait. Biomechanical analyses of stair-gait largely focus on lower extremity kinematics and kinetics with little attention on the whole body. Additionally, a great proportion of the work done in trunk kinematics has been in the frontal plane and in level gait. The trunk comprises of a large proportion of body mass and thus small changes in upper body motion could lead to considerable changes in lower extremity biomechanics. Therefore, characterizing trunk biomechanics in the sagittal plane during stair-gait is important in furthering our understanding of how the trunk is involved in movement strategies and how this may impact mobility task performance.
- 3 Only a few studies have investigated knee extensor capacity utilization during stair-gait. Currently, no study has investigated if knee extensor capacity utilization is linked to trunk kinematics in any respect. It is unclear to what extent one’s ability to effectively use their available knee extensor capacity may influence, or is influenced by, sagittal plane trunk kinematics. Therefore, exploring any potential relationship between these two factors will help discern whether particular compensatory strategies exist to negotiate stairs.

5 Objectives and Hypotheses

The purpose of this investigation is to assess sagittal trunk motion during stair-gait and evaluate how different knee extensor capacities might impact movement strategies in younger and older adults. This investigation will hopefully identify biomechanical features of stair-gait in people of differing ages well as the potential impact of knee extensor muscle function on performance.

5.1 Objectives

- (1) To determine if knee extensor capacity utilization differs between younger and older adults for both stair ascent and descent, respectively.
- (2) To determine if a relationship exists between knee extensor capacity utilization and sagittal trunk kinematics for stair ascent and descent, respectively.

5.2 Hypotheses

Older adults have lower knee extensor muscle strength while also exhibiting greater knee extensor capacity utilization during various ADLs; few studies demonstrate this during stair ascent and descent. Older adults are expected to demonstrate higher knee extensor capacity utilization than younger adults during stair ascent and descent. It is hypothesized that both older and younger adults will exhibit different knee extensor capacity utilizations for both stair ascent and descent (Objective 1). Due to the proposed mechanism by which increased trunk flexion may reduce knee joint demands, it is hypothesized that knee extensor capacity utilization will be associated with sagittal trunk kinematics, where those working at higher capacity utilization will demonstrate greater sagittal plane trunk flexion during stair-gait (Objective 2).

6 Methodology

6.1 Study Design

The current study is a single-visit, cross-sectional descriptive study on healthy young and older aged participants. Experimental procedures involved performance-based tests and questionnaires to quantify physical function and pain; maximal isometric and isokinetic contractions to estimate knee extensor muscle capacity; as well as kinematics and kinetics during stair ascent and descent on an instrumented, custom-built staircase composed of four steps. The study visit took 2.5-3 hours. Participants were remunerated \$25.00 in cash for their time.

6.2 Participants

6.2.1 Inclusion Criteria

Individuals were permitted to participate in this study if they were self-reported healthy individuals aged between 18-30 years or 45-70 years.

6.2.1.1 Exclusion Criteria

Participants were excluded if they met the American College of Rheumatology (ACR) clinical classification for unilateral and/or bilateral knee OA (Altman et al., 1986). The ACR classification criteria includes knee pain plus at least 3 of the following 6 clinical signs or symptoms: i) be 50 years or older; ii) morning stiffness for less than 30 minutes; iii) crepitus with active motion; iv) bony tenderness upon palpating margins of joint; v) bony enlargement; and vi) no signs of palpable warmth of the synovium (Altman et al., 1986). Also, participants were excluded if they self-reported any of the following: over 70 years of age, an unstable heart condition, alternate forms of arthritis, history of osteoporotic fractures, active non-arthritis knee disease (e.g., gout, rheumatoid arthritis), persistent use of a walking aid, neurological conditions, previous knee surgery or current intraarticular therapy, an injury to the lower limbs in the past three months, spinal, hip, and/or ankle conditions, current cancer treatment, pregnancy, or unable to independently ascend and descend a set of stairs. Potential participants were excluded if their

measured height and body mass, while wearing a t-shirt, shorts and barefoot, computed a body mass index (BMI) $\geq 36 \text{ kg/m}^2$. Finally, a score from the Get Active Questionnaire (GAQ) indicating a health condition may limit exercise participation, was an exclusion criterion.

6.2.2 Recruitment Strategy

Participants were recruited through posters, social media, and word-of-mouth in the Waterloo, Ontario area. Posters were placed around the University of Waterloo campus including the Center for Community, Clinical and Applied Research Excellence (CCCARE). Digital recruitment strategies were utilized through the University of Waterloo Kinesiology and Health Sciences Graduate Community Group Portal. Participants interested in joining the study contacted investigators via telephone or email. When a potential participant contacted a member of the research team, they were informed of study details and screened if they met inclusion and exclusion criteria as well as any COVID-19 safety standards at the time of the study.

6.2.3 Sample Size Estimation

A sample size estimate was performed using data from the literature which both included robust sample size, while also estimated and compared muscle capacity utilization across two groups similarly (Samuel et al., 2011). Samuel et al. (2011) compared muscle capacity utilization between males and females for both stair ascent and descent in a sample of 84 healthy older adults. The authors used a linear interpolation method to estimate muscle capacity and subsequently calculate muscle capacity utilization (Samuel et al., 2011). They also reported differences between males [$n = 41$ (mean $\pm$ SD age: 73.4 ± 7.1 years; height: 173.0 ± 7.0 cm; body mass: 78.7 ± 13.7 kg)] and females [$n = 43$ (mean $\pm$ SD age: 72.9 ± 7.6 years; height: 160.0 ± 7.0 cm; body mass: 69.1 ± 13.4 kg)] for both stair directions (Samuel et al., 2011). During stair ascent, the mean knee extensor functional demand (%) was 93.8 ± 36.6 for males and 111.4 ± 40.0 for females (Samuel et al., 2011). During descent, the mean knee extensor functional demand (%) was 112.2 ± 46.7 for males and 129.2 ± 41.0 for females (Samuel et al., 2011). This data was used to

calculate effect sizes for sample size estimates. The sample size estimation was calculated using G*Power (v.3.1.9.7) software (Faul et al., 2009; Franz et al., 2007). A p-value of 5% and power of 80% have been accepted and used for this sample size calculation (Bhalerao & Kadam, 2010). This calculation yielded an estimated sample size of at least 18 males and 18 females for a total of 36 participants.

6.3 Experimental Protocol

Individuals who showed an interest in participating were emailed a consent form and a screening questionnaire including the GAQ. Potential participants were then contacted by a researcher via phone or email to address any questions regarding the study, to confirm study eligibility, and to schedule a session at the Mobilize Lab. An email reminder was sent to the participant at least 2 days prior with details on where to meet with a researcher, what to bring, and what to expect. Written informed consent was collected from eligible participants prior to initiating any experimental procedures. A researcher provided a brief explanation of what to expect from the study as well as an opportunity to raise any questions or concerns. At this point, verbal consent to participate was also obtained from the participant and were simultaneously informed that they were able to withdraw their consent at any point during the study.

The participant's age, height, body mass, and resting heart rate were first collected. The dominant leg, defined as the leg a participant would use to kick a ball or use to step forward to stop a fall, was defined as the study leg. Waist, hip, dominant thigh, and dominant calf circumference were measured as well as dominant knee width on the study leg. Participants then performed physical functional tests including the 6-Minute Walk Test (6MWT) and 9-Step Stair Climb Test (9SCT) as descriptors.

PROCEDURES	Screening & Consent	Anthropometrics & Descriptors	Physical Functional Tests	Dynamometry	Stair Ascent & Descent	Exit
MEASURES		Age, Height, Mass, Circumferences, etc.	6MWT 9SCT	Iso-metric Iso-kinetic	Kinematics & Kinetics	
QUESTIONNAIRES	BMI GAQ	NPRS (Pre) KOOS	Sex & Gender Race & Ethnicity	ABC-16		NPRS (Post) PD-Q

Figure 5.1: Outlined experimental protocol and associated measures and questionnaires. BMI = Body Mass Index. GAQ = Get Active Questionnaire. NPRS = Numeric Pain Rating Scale. KOOS = Knee Injury and Osteoarthritis Outcome Scale. 6MWT = 6-Minute Walk Test. 9SCT = 9-Step Stair Climb Test, ABC-16 = Activities-based Balance Confidence Scale, PD-Q = painDETECT Questionnaire.

6.4 Staircase Set-up and Stepping Protocol

Three-dimensional kinematics and kinetics of the trunk and lower extremity joints on the study limb were captured during stair ascent and descent as key variables of interest in this study. The following describes the custom staircase designed and built for the conduct of this study, the protocol completed by participants and the instrumentation used to capture kinematics and kinetics. Finally, this section describes how knee extensor muscle capacity utilization was calculated.

6.4.1 Stair Specifications

I led the design and observed the construction of a custom-built instrumented staircase consisting of 4 steps, each with a height of 17 cm, tread depth of 26 cm, and tread width of 100 cm. The last step doubled as a landing with a depth of 200 cm. A force plate (OR6-7, Advanced Mechanical Technologies Inc, Watertown, MA) was embedded, flush, and level with the middle of the second step. A single adjustable handrail with a height of ~85-95 cm relative to the treads followed the steps on the participants' non-dominant side to the landing which will have an additional connecting handrail at the end (Figure 5.2). This handrail set-up was intended to minimize potential marker obstruction. Stair parameters are

based on the most recent Ontario building code to maximize transferability of task performance from participants' daily lives (Building Code Act, 1992).

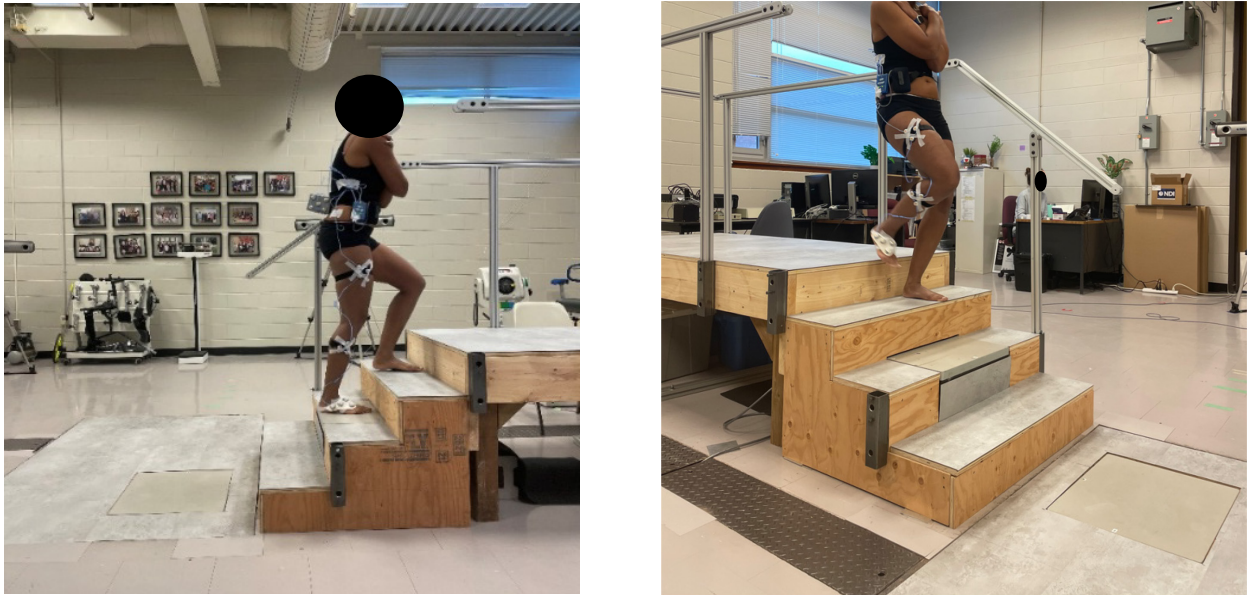


Figure 5.2: Staircase set-up for stair-ascent (left) and stair descent (right) for a right leg dominant individual.

6.4.2 Stepping Protocol

Participants were asked to take a 2-3 step walking start towards the stairs, then transition to stair ascent. Participants used a step-over-step strategy while ensuring complete foot contacts on both force plates. The arm on the participant's study leg side was positioned across the chest to minimize potential marker obstruction. The contralateral arm was able to move freely. Participants were permitted to use the handrail if they felt they may lose balance but were discouraged to do so otherwise. If the participant contacted the handrail, the trial was considered unsuccessful and removed. Participants completed 3 practice trials to get acquainted with the stair ascent task prior to data collection. Similar procedures were completed for stair descent. A total of 10 trials were collected for both stair ascent and descent.

6.5 Kinematics

Three-dimensional kinematics were captured using a 5-camera bank (15 cameras) high-speed motion capture system (Optotrak Certus, Northern Digital Inc., Waterloo, ON, CA) sampled at 100 Hz (Longpré, et al., 2015). Calibration and registration of the collection volume was completed using a calibration cube with a known marker alignment. The calibration cube was then placed on the upper-right corner of a floor-embedded force plate facing the stairs to define the origin of the global coordinate system with +X pointing orthogonal and to the left of +Y which is pointing in the direction of progression during stair ascent, or towards the staircase, and +Z pointing downwards. Five rigid body clusters containing triad-oriented infrared light emitting diodes (iRED) were affixed to the participants' cervical spine, superior sacrum, as well as unilaterally on the mid-lateral thigh, mid-lateral shank, and antero-lateral aspect of the foot on the dominant leg. Rigid-linked bodies for the lower extremity (i.e., foot, shank, and thigh), pelvis, and trunk segments were created and defined using digitized anatomical landmark coordinates found in Table 1 (Brisson et al., 2018; de Leva, 1996; Robertson et al., 2014; Wu et al., 2002). Collectively, these formed a 5-segment rigid-linked model for each participant. A five second static standing pose trial was used to transform the 5-segment external markers to a reference frame in processing. A 20 second dynamic trial of repeated knee flexion and extension with the hip in extension was used to calibrate and define a functional knee joint center. Additionally, a 20 second dynamic trial was used including a combination of counterclockwise and clockwise hip circumduction as well as creating a star-like pattern pointing to 0, 45, 90, 135, and 180 degrees with the leg in a straightened position. This was used to calibrate and define a functional hip joint center.

Table 5.1: Digitized anatomical landmark locations for the foot, shank, and thigh with their respective rigid body cluster.

Marker	Anatomical Landmark	Rigid Body Cluster
1	8 th Thoracic Vertebral Spinous Process	Cervical Spine
2	7 th Cervical Vertebral Spinous Process	Cervical Spine
3	(R) Acromion Process	Cervical Spine
4	(L) Acromion Process	Cervical Spine
5	Suprasternal Notch	Cervical Spine
6	Xiphoid Process	Cervical Spine
7	(R) Iliac Crest	Sacrum
8	(L) Iliac Crest	Sacrum
9	(R) Anterior Superior Iliac Spine	Sacrum
10	(L) Anterior Superior Iliac Spine	Sacrum
11	(R) Posterior Superior Iliac Spine	Sacrum
12	(L) Posterior Superior Iliac Spine	Sacrum
13	Greater Trochanter	Thigh
14	Lateral Epicondyle	Thigh
15	Medial Epicondyle	Thigh
16	Fibular Head	Shank
17	Tibial Tuberosity	Shank
18	Lateral Malleolus	Shank
19	Medial Malleolus	Shank
20	Calcaneus/ Achilles Tendon Insertion	Foot
21	Dorsal Base of 5 th Metatarsal Head	Foot
22	Dorsal Base of 2 nd Metatarsal Head	Foot
23	Dorsal Base of 1 st Metatarsal Head	Foot

L = Left Side

R = Right Side

The Optotrak Certus has demonstrated to be both a precise and reliable measurement tool for the collection of waveform data in gait analyses (Mazumder et al., 2007). Mazumder and colleagues (2007) used a 3D robot to test the precision and repeatability of the Optotrak Certus. They found excellent precision for between marker distance (static: % error = ~0.57; SD = ~0.003; dynamic: % error = ~0.575; SD = ~0.007) and between marker angles (% error = ~4.01; SD = ~0.000163) as well as excellent repeatability for angle (coefficient of multiple correlation (CMC) = 0.9901; CV (%) = 1.2634; SD = 1.0917), volume (CMC = 0.9999; CV (%) = 1.4151; SD = 0.0016), and circular motion (x-axis: CMC = 0.9949; CV (%) = 2.8104; SD = 0.00944; y-axis: CMC = 0.9956; CV (%) = 0.7761; SD = 0.008266) (Mazumder et al., 2011).

6.6 Kinetics

Kinetic data (ground reaction forces) were collected synchronously with the motion capture system using floor-embedded force plate and step-embedded force plate (OR6-5 & OR6-7, respectively, Advanced Mechanical Technologies Inc, Watertown, MA, USA) sampled at 1000 Hz (Brisson et al., 2018). Kinetics were collected on the dominant leg. Force plates were turned on at least 45 minutes prior to collections to minimize drift during the data collection. Force plates were zeroed before collections and between each stair-gait trial.

6.7 Muscle Capacity

Muscle capacity (strength and power) was measured with a commercial strength dynamometer (System 4 Pro, Biodex Medical Systems Inc, Shirley, New York) using a procedure consistent with the manufacturer's recommendations. Prior to recording, participants were given a brief overview of testing protocols. If any pain, discomfort, or other adverse events were to occur during testing, the participant was informed to state this to the researcher immediately. Participants were then positioned at 90 degrees of trunk flexion and secured by straps at the pelvis, dominant thigh, and upper torso. The leg attachment cuff secured the dominant shank 2-3 cm above the lateral malleolus. Participants were informed that straps were to be secured tightly but should not be too tight to cause discomfort. The knee joint center was aligned with the rotational axis of the dynamometer. The lower limb was weight and gravitation corrected prior to both isometric trials and isokinetic trials. Before each testing protocol, participants underwent a submaximal warm-up consisting of repetitive, submaximal knee flexion and extension efforts.

Every participant underwent a standardized protocol to measure the MVIC for the knee extensors of the dominant leg at 60° of knee flexion (Accettura et al., 2015). Each trial required the participant to hold an MVIC for five seconds followed by sixty seconds of rest; this procedure was repeated five times (Accettura et al., 2015). Verbal encouragement and visual aid (i.e., torque produced in real time) were provided during contractions. Raw time, torque, and angular velocity data from the five isometric trials were extracted for processing.

Maximal isokinetic knee extensor torque was measured using a standardized protocol at the following angular velocities: 60 deg/s, 90 deg/s, and 180 deg/s, both concentrically and eccentrically. A maximum velocity of 180 deg/s was chosen as 270 deg/s demonstrated to reduce the potential sampling window as there was not enough range of motion available to reach and sustain this velocity and raised concerns over the validity of measuring capacity at this speed. Each test was conducted by alternating between concentric and eccentric contractions through a total available knee range of motion of approximately 70° for 3 sets of 3 repetitions for a total of 9 concentric and 9 eccentric repetitions at each velocity (Hafer & Boyer, 2020). Verbal encouragement and visual aid were provided during these contractions. Participants were given 60 seconds rest between sets to minimize the impact of fatigue on performance (Hafer & Boyer, 2020). Raw time, torque, and angular velocity data from the 3 testing velocities for each contraction type were extracted for processing.

Commercial dynamometers have been used extensively in research and clinical environments for the quantification of muscle capacity and function (van Driessche et al., 2018). Studies have demonstrated both the Biodex System 3 and System 4 dynamometer both to yield reliable and valid data on knee extensor and flexor strength and power (Brisson et al., 2018; Drouin et al., 2004; van Driessche et al., 2018). Drouin et al. (2004) demonstrated near perfect trial-to-trial and day-to-day reliability and validity for the measurement of angular position, isometric torque, and slow to moderately high angular velocities up to 300 deg/s using the Biodex System 3. Another study investigated the relative and absolute test-retest reliability of rapid knee extension in older adults using the isotonic mode on a Biodex System 3 dynamometer and found high test-retest reliability for relatively low loads (i.e., 0%, 25%, & 50% MVIC) (Drouin et al., 2004; van Driessche et al., 2018). However, poor reliability was found at increasing loads (i.e., 75% MVIC) and within the acceleration phase (i.e., first 50 ms) likely the result of poor performance and differences in neural activation characteristics, respectively (Drouin et al., 2004; van Driessche et al., 2018). Dynamometry produces reliable data among individuals with knee OA for the measurement of

quadriceps strength (Intraclass Correlation Coefficient (ICC) = 0.91 in N/kg; ICC = 0.93 in N) and power (ICC = 0.84 in N/kg; ICC = 0.88 in N) (Brisson et al., 2018).

6.8 Performance-based Tests

Standard measurements of physical performance used in clinical practice were measured to describe the sample. The 6MWT (See Appendix F) was used to assess gait speed. Participants were asked to walk at a self-selected speed as far as possible within six minutes in a safe manner (Dobson et al., 2013). A 39.2 metre rectangular circuit was used to travel a track distance. Participants could not jog or run but were permitted to slow down, sit, or stop if needed; the timer continued regardless (Dobson et al., 2013). The 6MWT has demonstrated excellent test-retest reliability in people with primary (ICC = 0.991, 95% CI = 0.986-0.994) and end-stage knee OA (ICC = 0.94, 95% CI = 0.88, 0.98) as well as good content and concurrent validity (Ateef et al., 2016; Kennedy et al., 2005).

The 9-SCT was used to determine stair-gait speed. Participants were asked to ascend and descend a set of 9 steps as quickly and safely as possible (See Appendix G) (Dobson et al., 2013). Participants started with both feet on the ground at the first step and were timed from initiation of movement until both feet have returned to the ground at the last step (Dobson et al., 2013). Step strategy, step-over-step or step-by-step, as well as handrail contact was documented (Dobson et al., 2013).

6.9 Questionnaires

6.9.1 Physical Function

Physical function was quantified to describe the sample. Prior to performance-based tests, knee symptom severity and function were quantified using the Knee injury and Osteoarthritis and Outcome Scale (KOOS) (See Appendix D). The KOOS is a 42-item self-report questionnaire evaluating short and long-term knee-related outcomes in people with knee injury and/or knee OA using five scoring subscales: pain, symptoms, activities of daily living function, sport and recreation function, and knee-related quality of life (Collins et al., 2016; Roos et al., 1998; Roos & Lohmander, 2003). Each outcome variable is

quantified using a standardized five-item Likert scale scored from 0 (No Problems) to 4 (Extreme Problems) (Roos et al., 1998; Roos & Lohmander, 2003). The scores were summed and converted to a 0-100 scale with 0 indicating worst possible knee symptoms and 100 indicating no knee symptoms (Roos et al., 1998; Roos & Lohmander, 2003). Each subscale for the KOOS has demonstrated adequate test-retest reliability (ICC range: 0.85-0.90; $I^2 \geq 68.9\%$) as well as moderate construct validity when compared to SF-36 general health status subscales (physical function vs. ADL, $r = 0.57$; physical function versus sport and recreation function, $r = 0.47$; bodily pain vs. pain, $r = 0.46$) (Collins et al., 2016; Roos et al., 1998).

Self-reported confidence with various activities was collected to describe the sample. The Activity-specific Balance Confidence (ABC-16) scale contains 16 items to assess balance confidence during ADLs including: walking around the house; walking up and down stairs; bending over to pick slippers up from floor; reaching for a small can on a shelf at eye-level; reaching above head on tiptoes; standing on chair to reach; sweeping the floor; walk outside to nearby car; getting into/out of car; walking across a parking lot; walking up or down a ramp; walking through a crowded mall where people rapidly walk past you; bumping into people in a crowd; walking onto an escalator holding rail; walking onto escalator not holding rail; walking outside on icy sidewalks (see Appendix) (Powell & Myers, 1995; Schepens et al., 2010). Participants rated their balance confidence for each item on the ABC-16 scale on a continuum from 0-100 (Powell & Myers, 1995; Schepens et al., 2010). The scores on the 16 items were averaged and represented as a single percentage score with 0 indicating no confidence and 100 indicating complete confidence (Powell & Myers, 1995; Schepens et al., 2010). The ABC-16 produces data with high reliability ($r = 0.96$) and good internal consistency (Cronbach's $\alpha = 0.96$) in community-dwelling older adults (Powell & Myers, 1995).

6.9.1 Self-reported Pain

Dimensions of pain were captured to describe the sample. Participants self-reported their pain on the Numeric Pain Rating Scale (NPRS) at the beginning and at the end of the testing session separately for

each limb. The NPRS rates participants' current perceived pain on a scale of 0-10 with 0 being no pain and 10 being immense pain. Excellent test-retest reliability of the NPRS (ICC = 0.95) as well as good-to-excellent correlation with the Visual Analog Scale (VAS) ($r = 0.941$) has been demonstrated for pain ratings in middle-aged to older adults with knee OA (Alghadir et al., 2018).

At the end of the testing session, participants were provided with the painDETECT questionnaire (PD-Q), which is a self-report screening tool designed, initially for chronic low back pain, to distinguish between nociceptive and neuropathic pain (Freynhagen et al., 2006) (See Appendix C). Pain gradation includes seven questions scored on a six-item Likert scale from 0 (never) to 5 (very strongly); pain course pattern uses images depicting a pattern that is associated with a numeric score (-1 to 1); and the presence of radiating pain using a body map as well as a score of 2 if present and 0 if not present (Freynhagen et al., 2006). Total score ranges from -1 to 38 (Freynhagen et al., 2006). Several cut-offs are used for screening: score ≤ 12 unlikely neuropathic-like pain ($<15\%$); score 13-18 uncertain if neuropathic-like pain present; score ≥ 19 likely neuropathic-like pain ($>90\%$) (Freynhagen et al., 2006). The PD-Q has demonstrated moderate test-retest reliability in people with knee OA (ICC = 0.58; 95% CI = 0.36, 0.76) (Turner et al., 2017). The PD-Q has good content and criterion validity against expert pain physician diagnosis in a large sample of patients while also demonstrating 84% sensitivity and specificity (Freynhagen et al., 2006).

6.10 Data Processing

6.10.1 Kinematics & Kinetics of Stair Ascent and Descent

Pose and stair-gait kinematic and kinetic data were processed using commercial biomechanics software (Visual3D, C-Motion, Germantown, MD). Of the ten trials collected for each of ascent and descent, a maximum of 7 trials were selected for analysis. Kinematic and kinetic was filtered using a low-pass fourth-order Butterworth filter with a 6 Hz cut-off frequency and interpolated using a third-order polynomial with a maximum gap of 10 frames (Brisson et al., 2018; Hicks-Little et al., 2011; Longpré et

al., 2015; Winter, 2009). One full stair-gait cycle was defined as the time from one toe-off to the next ipsilateral toe-off and time-normalized to 100% stair-gait. For stair ascent, a stair-gait cycle was further segmented into a stance phase with 3 additional sub-phases (Weight Acceptance, Pull-up, and Forward Continuance) and a swing phase with 2 sub-phases (Foot Clearance and Foot Placement) (Table 2.2). For stair descent, a stair-gait cycle was further segmented into a stance phase with 3 additional sub-phases (Weight Acceptance, Pull-up, and Forward Continuance) and a swing phase with 2 sub-phases (Foot Clearance and Foot Placement Preparation) (Table 2.3).

Body segment inertial parameters (mass, center of mass, and mass moment of inertia) were calculated for each participant using anatomical measurements and regression models (McGibbon Puniello, & Krebs, 2001). Kinematic, kinetic, and body segment inertial parameters were input into Newton-Euler inverse dynamic equations to calculate the net joint reaction forces (N) and net internal joint moments (Nm) at the ankle, knee, and hip for 100% ascent and descent stance phases McGibbon Puniello, & Krebs, 2001; McGibbon & Krebs, 2001; Segal, Yack, et al., 2009).

Table 5.2: Vertical ground reaction force (vGRF) expressed relative to body weight (BW) and centre of mass (COM) were used to define phases and subphases of stair ascent.

Phase	Sub-Phase	Sequence	Definitions
Stance	Weight Acceptance (WA)	IC → SLS	Initial Contact (IC): Instance where vGRF exceeds 1% of participants BW for first time.
	Pull-Up (PU)	SLS → MST	Single-leg Stance (SLS): Instance where vGRF equals 100% participant's BW for first time.
	Forward Continuance (FWC)	MST → TO	Mid-stance (MST): Instance where the whole-body COM is directly above (i.e., crossing over) the stance foot COM. Net distance in y-directed COM is zero. Toe-off (TO): Instance where vGRF crosses zero N.
Swing	Foot Clearance (FCL)	TO → MSW	Mid-swing (MSW): Instance where the swing foot COM is directly below (i.e., crossing under) the whole-body COM. Net distance in y-directed COM is zero.
	Foot Placement Preparation (FPP)	MSW → LSW	Late-swing (LSW): Instance where the swing foot prepares for next contact just before next ipsilateral foot contact and reaches peak vertical velocity.

vGRF = vertical ground reaction force.
COM = center of mass.

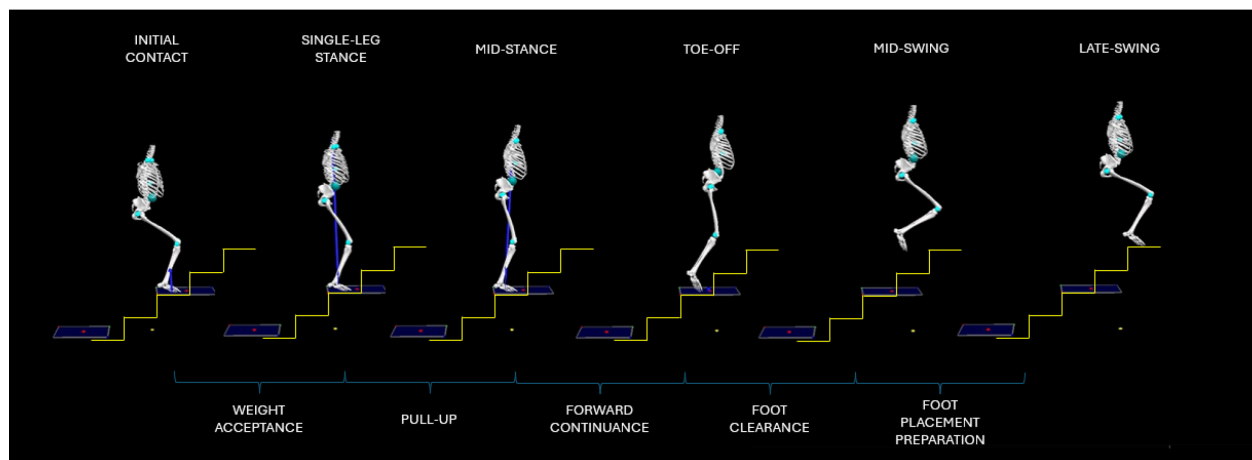


Figure 5.3: Skeletal model of the different phases of stair ascent from commercial biomechanics software (Visual3D, C-Motion, Germantown, MD).

Table 5.3: Vertical ground reaction force (vGRF) expressed relative to body weight (BW) and centre of mass (COM) were used to define phases and subphases of stair descent.

Phase	Sub-Phase	Sequence	Definitions
Stance	Weight Acceptance (WA)	IC → SLS	Initial Contact (IC): Instance where vGRF exceeds 1% of participants BW for first time.
	Forward Continuance (FWC)	SLS → MST	Single-leg Stance (SLS): Instance where vGRF equals 100% participant's BW for first time.
	Controlled Lowering (CTL)	MST → TO	Mid-stance (MST): Instance where the whole-body COM is directly above (i.e., crossing over) the stance foot COM. Net distance in y-directed COM is zero. Toe-off (TO): Instance where vGRF crosses zero N.
Swing	Leg Pull-through (LPT)	TO → MSW	Mid-swing (MSW): Instance where the swing foot COM is directly below (i.e., crossing under) the whole-body COM. Net distance in y-directed COM is zero.
	Foot Placement Preparation (FPP)	MSW → LSW	Late-swing (LSW): Instance where the swing foot prepares for next contact just before next ipsilateral foot contact and reaches minimum vertical velocity.

vGRF = vertical ground reaction force.
COM = center of mass.

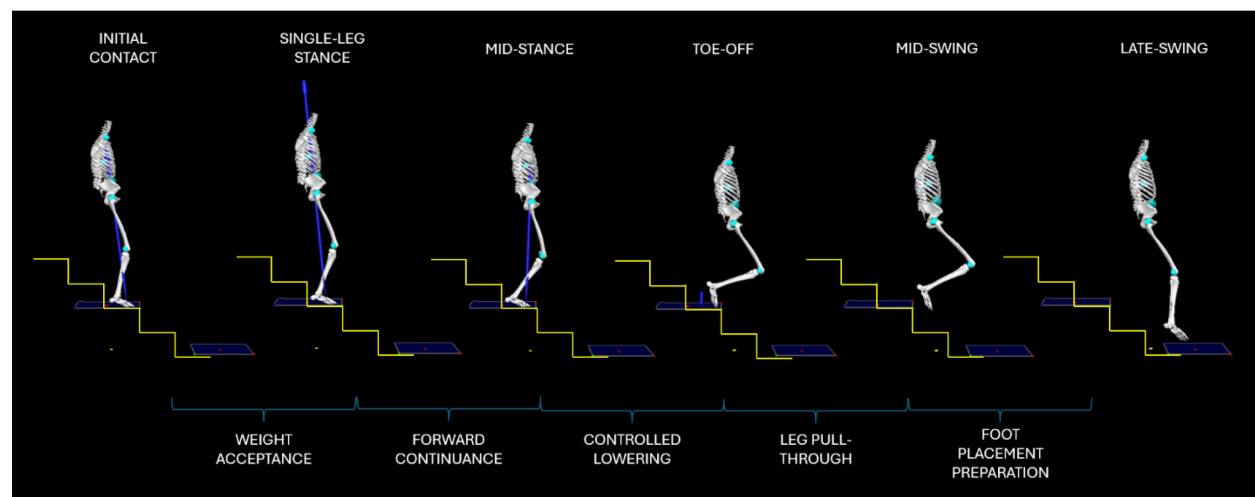


Figure 5.4: Skeletal model of the different phases of stair descent from commercial biomechanics software (Visual3D, C-Motion, Germantown, MD).

Several assumptions are being made regarding the skeletal structures being modelled. First, they are represented as frictionless joints comprised from rigid segments which do not accurately reflect the dynamic and complex bone structures of several segments (i.e., the foot and trunk) (Robertson et al., 2014). This approach allows for manageable mathematical processing and definitions of Local Coordinate Systems (Robertson et al., 2014). Additionally, errors stemming from motion capture data (i.e., soft tissue artifact) are amplified when differentiating to calculate velocities and accelerations (Robertson et al., 2014; Winter, 2009). To mitigate these sources of error, standardized procedures involving marker placement over the skin and bony landmark digitization were carefully followed.

The global (laboratory) coordinate system was defined as a right-handed orthogonal system, with the +Z axis oriented vertically and aligned with the direction of gravitational acceleration, the +Y axis oriented anteriorly along the direction of progression during stair ascent, and the +X axis oriented laterally to the left, orthogonal to both +Y and +Z. Local segmental coordinate systems were also defined as right-handed orthogonal triads, with the Z-axis aligned with the segment's longitudinal axis, the Y-axis directed anteroposteriorly, and the X-axis oriented mediolaterally. These conventions, while biomechanically consistent, do not strictly follow ISB recommendations (Wu et al., 2002; Wu & Cavanagh, 1995). Floating three-dimensional axis model was used to determine the relative angular motion between adjoined segments (i.e., joint angles) (Grood & Suntay, 1983; Wu et al., 2002). Joint kinematics (i.e., angles, velocities, and accelerations) and kinetics (i.e., moments) for the ankle, knee, and hip were calculated using an X-Y-Z Cardan rotation sequence using commercial software (Visual 3D, C-Motion, Inc., Germantown, MD, USA). Trunk flexion/extension was defined as the angle formed by a line running rostrally from the mid-thoracic to the T8 landmarks and the global vertical axis.

6.10.2 Knee Extensor Capacity & Utilization

A custom Python (version 3.11.4) script was used to calculate knee extensor capacity and utilization from the raw isometric and isokinetic data. Of the 5 MVIC efforts, the peak torque from the 3 trials with

the greatest peak torque magnitudes were averaged and normalized to body mass (Nm/kg) to represent a single maximal knee extensor strength value for each participant (Accettura et al., 2015; Brisson et al., 2018; Segal et al., 2013) (Table 2). Isokinetic data were windowed to contain only velocities within ± 1 deg/s of the set velocity for each repetition. This windowing was done to remove data that included acceleration and deceleration phases. During the concentric contractions set to 180 deg/s, three participants were unable to reach 180 deg/s concentrically and instead reached 175.4 deg/s, 174.5 deg/s, and 174.5 deg/s, respectively. These data were used for analysis. The 3 greatest peak torque magnitudes from any of the 3 sets were averaged and normalized to body mass (Nm/kg) to yield a single maximal knee extensor isokinetic value at each speed for a total of 6 metrics: 60, 90, and 180, deg/s concentrically as well as 60, 90, and 180 deg/s eccentrically. Maximal knee extensor isometric and isokinetic torque values were fit with a third-order polynomial to create a participant-specific moment-velocity curve (Hafer & Boyer, 2020).

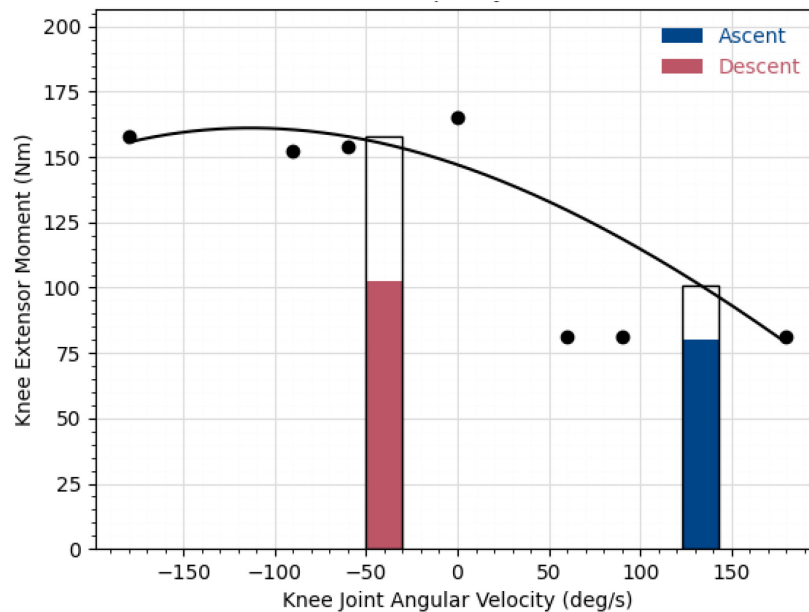


Figure 5.5: Visual representation of a participant's knee extensor utilization. Dots represent maximal isometric and isokinetic knee moments at 60, 90, and 180 deg/s both concentrically (+) and eccentrically (-) fit with a third-order polynomial. Coloured portion of bars represent peak external knee flexion moment at the angular velocity this was achieved for both ascent and descent. Bar limits represent estimated knee extensor capacity (i.e. y-value on third-order polynomial corresponding with angular velocity at peak external knee flexion moment).

Operationally, knee extensor capacity utilization (KECU) was defined as the quotient (%) between the peak EKFM (Nm/kg) observed during stance and the estimated knee extensor capacity (KEC). In this study, KECU was calculated using methods described by Hafer & Boyer (2020) for both ascent and descent separately. This approach calculates KEC (denominator) by estimating the knee extensor moment on the participant-specific moment-velocity third-order polynomial corresponding with the angular velocity (°/s) at which the peak EKFM occurred during stair-gait (Equation 1). Previous studies have also calculated KECU by using only the MVIC as the denominator (Equation 2) (Brisson et al., 2018; Hortobágyi et al., 2003; Tung et al., 2022). There were 2 healthy young participants that achieved a peak EKFM beyond what was measured isokinetically; the MVIC was used to compute KECU for these participants (Equation 2). During stair descent, there were 9 participants (6 HYA, 3 HOA) that achieved peak EKFM at angular velocities beyond what was captured via eccentric isokinetic dynamometry; the MVIC was used to compute KECU for these participants (Equation 2).

$$(1) \quad KECU (\%) = \frac{\text{Peak EKFM}_{\text{stair-gait}} (\text{Nm/kg})}{KEC}$$

$$(2) \quad KECU (\%) = \frac{\text{Peak EKFM}_{\text{stair-gait}} (\text{Nm/kg})}{MVIC}$$

6.11 Statistical Analyses

Two independent analyses were conducted to obtain descriptive statistics (means, medians, standard deviations, and ranges) for stair ascent and descent, respectively. Prior to these statistical tests, the Shapiro-Wilk test was used to determine if data were normally distributed and Levene's test was used to determine if there was any violation to the assumption of homogeneity of variance. In the event a parametric statistical test was not appropriate, a non-parametric statistical test was used.

As there was no directionality to objective (1), a two-tailed Welch's t-test was used to determine if there was any difference in knee extensor capacity utilization between the healthy young adults and healthy older adults for stair ascent; this was repeated for stair descent. Effect sizes were calculated using Cohen's d. To approach research objective (2), a Pearson-Product Moment Correlation (PPMC) test was

used to determine the relationship between knee extensor capacity utilization and the peak trunk flexion angle during the stance phase of stair-gait; this was repeated for stair descent.

7 Results

7.1 Participants

Twenty-five healthy young adults aged 18-30 years and healthy older adults aged 45-70 years were recruited primarily from the University of Waterloo and surrounding communities. The healthy young adult group contained more racial diversity than the healthy older adult group, which is likely a reflection of the University of Waterloo student population relative to the surrounding community (Table 3.1). BMI and Waist-to-Hip Circumference Ratio in healthy older adults was larger than in healthy young adults (Table 3.2).

Table 6.1: Participant demographics.

		Healthy Young Adult (HYA) (n = 16)		Healthy Older Adult (HOA) (n = 9)	
		n	%	n	%
Sex	Female	9	56.3	4	44.4
	Male	7	43.7	5	55.6
Gender	Woman	9	56.3	4	44.4
	Man	7	43.7	5	55.6
Race*	Black	1	5.6	0	0
	Middle Eastern	3	16.7	0	0
	East Asian	4	22.2	0	0
	South Asian	2	11.1	1	11.1
	Southeast Asian	2	11.1	0	0
	White	6	33.3	8	88.9

*Those reporting multiple races are presented here in the respective categories.

Table 6.2: Participant anthropometrics and descriptors.

	Healthy Young Adult (HYA) (n = 16)		Healthy Older Adult (HOA) (n = 9)	
	Mean (SD)	Range (Min-Max)	Mean (SD)	Range (Min-Max)
Age (y)	24 (4)	18 – 29	55 (8)	45 – 68
Height (m)	1.70 (0.09)	1.53 – 1.84	1.69 (0.09)	1.55 – 1.82
Body Mass (kg)	66.4 (15.1)	43.2 – 92.6	78.9 (21.8)	54.0 – 115.2
BMI (kg/m ²)	23.2 (3.40)	17.0 – 28.8	27.1 (4.8)	21.5 – 34.8
Waist:Hip Circumference	0.77 (0.10)	0.55 – 0.94	0.96 (0.23)	0.71 – 1.46
Knee Width (cm)	10.1 (0.6)	8.8 – 11.0	10.5 (1.1)	8.6 – 11.8

Abbreviations: Min, Minimum; Max, Maximum; BMI, Body Mass Index.

7.2 Descriptors

All participants reported pain scores that would reflect no, or very little, lower-extremity pain measured using the NPRS, PD-Q, and pain section of the KOOS (Table 3.3). HYA were able to walk a greater distance in 6 minutes compared to HOA. Additionally, HYA were able to ascend and descend the 9 steps in less time than the HOA.

Table 6.3: Sample descriptive characteristics.

		Healthy Young Adult (HYA) (n = 16)		Healthy Older Adult (HOA) (n = 9)	
		Mean (SD)	Range (Min-Max)	Mean (SD)	Range (Min-Max)
NPRS-Pre					
	Left Leg	0.39 (1.39)	0 – 5	0.67 (1.67)	0 – 5
	Right Leg	0.77 (0.28)	0 – 1	0 (0)	0 – 0
NPRS-Post					
	Left Leg	0.4 (1.4)	0 – 5	0.7 (1.1)	0 – 3
	Right Leg	2.0 (1.6)	0 – 5	0.7 (1.1)	0 – 3
6-MWT (m)		609.2 (57.3)	520.4 – 705.6	567.0 (83.7)	438.4 – 719.9
9-step SCT (s)					
	Ascent	2.71 (0.37)	2.09 – 3.38	3.43 (0.82)	2.40 – 4.75
	Descent	2.43 (0.41)	1.74 – 3.19	2.80 (0.67)	2.05 – 3.84
painDETECT		2.9 (4.3)	0 – 14	1.3 (2.0)	0 – 6
ABC-16		94.4 (6.4)	78.1 – 100	97.7 (1.8)	94.0 – 99.7
KOOS					
	Symptoms	96.4 (5.1)	85.7 – 100	97.2 (1.8)	92.9 – 100
	Pain	98.9 (1.8)	94.4 – 100	96.9 (6.3)	80.6 – 100
	Activities of Daily Living	99.3 (2.1)	92.7 – 100	98.9 (1.9)	95.6 – 100
	Sport & Recreation	95.0 (8.9)	75.0 – 100	94.4 (8.5)	75.0 – 100
	Knee Quality of Life	97.6 (5.4)	81.3 – 100	93.8 (8.3)	81.3 – 100

Abbreviations: NPRS, Numeric Pain Rating Scale; 6- MWT, Six Minute Walk Test; 9-step SCT, Nine Step Stair Climb Test; ABC-16, Activities-based Confidence-16 Items (Higher confidence with balance = 100); KOOS, Knee Injury and Osteoarthritis Outcome Score (Higher knee function = 100).

7.3 Dynamometry

Healthy older adults and healthy younger adults demonstrated relatively similar knee extensor capacity measured via isometric and isokinetic dynamometry (Table 3.4). On average, the highest peak knee extensor moments occurred in the isometric condition, and the lowest peak knee extensor moments occurred in the isokinetic concentric condition at 180 deg/s. Additionally, peak knee extensor moments in both concentric and eccentric isokinetic conditions appeared to decrease as speed increases. This is further illustrated in Figures 3.1 and 3.2. When accounting for body mass, relative knee extensor

moments are lower in healthy older adults compared to healthy young adults across all dynamometer test conditions.

Table 6.4: Absolute and relative maximal voluntary isometric knee extensor muscle contractions and maximal voluntary isokinetic knee extensor muscle contractions measured at 60, 90, 180 deg/s concentrically and eccentrically measured via dynamometry.

	Healthy Young Adult (HYA) (n = 16)		Healthy Older Adult (HOA) (n = 9)	
	Absolute (N·m)	Relative (N·m/kg)	Absolute (N·m)	Relative (N·m/kg)
Isometric	206.9 (77.9)	3.07 (0.69)	206.7 (81.8)	2.58 (0.47)
Isokinetic – Eccentric				
60 deg/s	188.9 (71.9)	2.84 (0.89)	201.1 (62.5)	2.56 (0.41)
90 deg/s	184.2 (74.9)	2.75 (0.86)	196.8 (56.8)	2.52 (0.38)
180 deg/s	163.7 (56.7)	2.48 (0.80)	165.1 (43.7)	2.14 (0.42)
Isokinetic – Concentric				
60 deg/s	139.7 (57.8)	2.08 (0.69)	143.9 (50.0)	1.82 (0.32)
90 deg/s	133.3 (55.7)	1.98 (0.59)	142.7 (38.2)	1.82 (0.19)
180 deg/s*	117.3 (56.2)	1.73 (0.66)	114.0 (34.6)	1.45 (0.23)

Data are presented as: Mean (SD).

*3 participants were unable to reach 180 deg/s. The peak knee extensor moment from the highest achieved velocity was used in place (178, 178, and 175 deg/s).

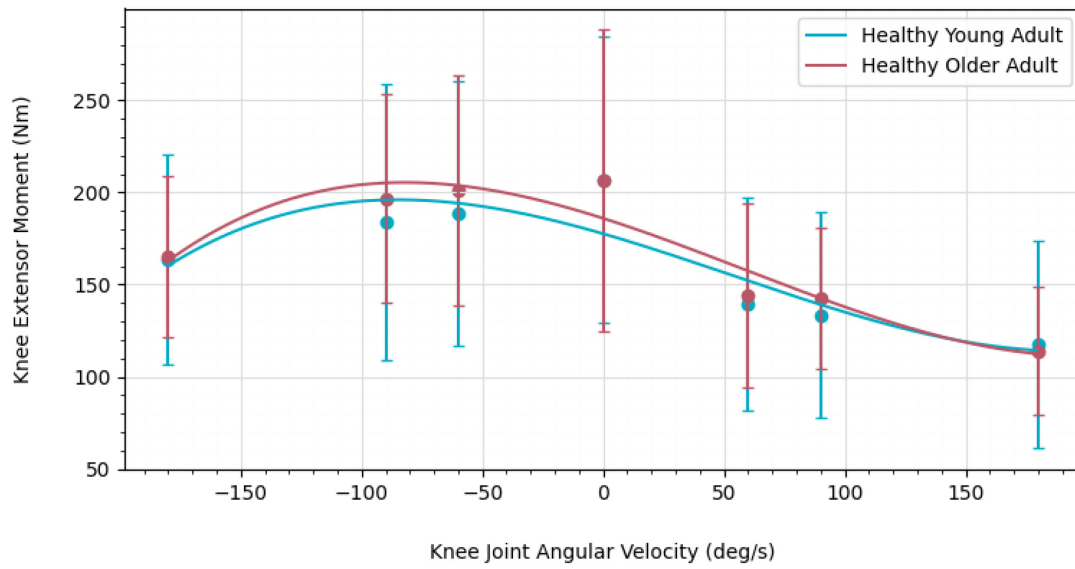


Figure 6.1: Absolute isometric and isokinetic knee extensor moments measured via dynamometry fitted with a third-order polynomial for each group.

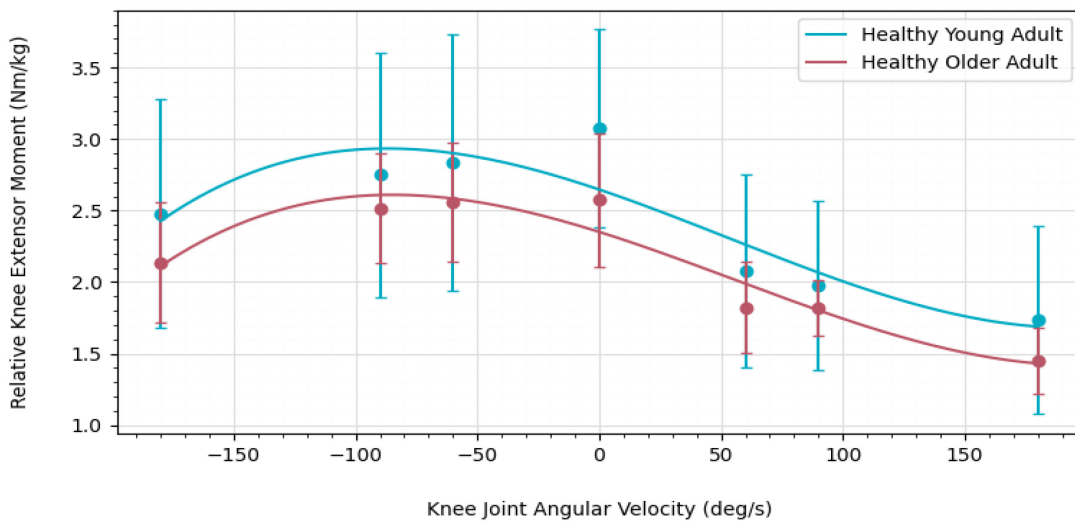


Figure 6.2: Relative isometric and isokinetic knee extensor moments (i.e., knee extensor moment divided by body mass) measured via dynamometry fitted with a third-order polynomial for each group.

7.4 Knee Extensor Capacity Utilization

While ascending stairs, healthy older adults used a higher KECU in comparison to healthy younger adults [$t(23) = 2.42, p = 0.02, d = -0.94, 95\% \text{ CI} = [-1.80, -0.08]$]. While descending stairs, healthy older and younger adults had similar KECU [$t(23) = 0.51, p = 0.62, d = -0.03, 95\% \text{ CI} = [-0.85, 0.79]$]. When observing the variables used to calculate KECU (Equations 1 and 2), both groups appear to have similar muscle capacities measured via dynamometry (see Tables 3.4 and 3.5). Similarly, the knee joint angular velocity at which peak EKFM occurred shows similarities between groups, although angular velocities appear slightly higher in the healthy young adult group during descent (Table 3.5). There are notably high standard deviations in knee extensor angular velocity during stair descent in both groups. Healthy older adults appear to demonstrate higher peak EKFM during both stair ascent and descent (Table 3.5).

Table 6.5: *Knee extensor capacity utilization and its constituents during both stair ascent and descent. Only KECU was statistically tested to match the primary research question a priori.*

	Ascent			Descent		
	HYA (n = 16)	HOA (n = 9)	<i>p</i>	HYA (n = 16)	HOA (n = 9)	<i>p</i>
KECU (%)	62.1 (13.8)	73.0 (8.8)	0.02*	57.2 (16.8)	57.6 (14.4)	0.62
Peak EKFM (N·m/kg)	1.26 (0.30)	1.19 (0.13)		1.48 (0.26)	1.34 (0.37)	
Estimated Muscle Capacity (N·m/kg)	2.06 (0.42)	1.64 (0.19)		2.94 (0.72)	2.44 (0.47)	
Peak EKFM (N·m)	84.6 (33.7)	92.8 (25.5)		98.2 (29.5)	103.1 (35.1)	
Estimated Muscle Capacity (N·m)	137.2 (45.7)	130.4 (44.0)		180.3 (51.2)	181.0 (51.8)	
Joint Velocity at Peak EKFM (deg/s)	114.1 (24.3)	117.5 (16.8)		-148.3 (93.0)	-132.8 (80.5)	

Abbreviations: HYA, Healthy Young Adult; HA, Healthy Adult; KECU, Knee Extensor Capacity Utilization; EKFM, External Knee Flexion Moment.

KECU Method:

Ascent – [KECU-POLY: HYA (14), HOA (9); KECU-ISOM: HYA (2), HOA (0)]

Descent – [KECU-POLY: HYA (7), HOA (6); KECU-ISOM: HYA (9), HOA (3)]

* = $p < 0.05$.

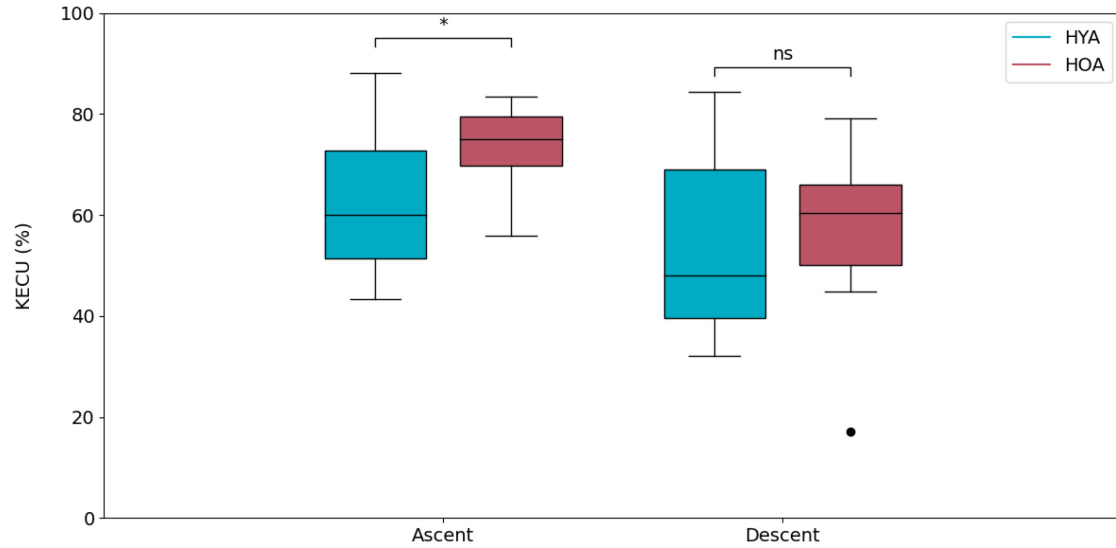


Figure 6.3: Figure 3.3: Knee extensor capacity utilization measured by groups for both ascent and descent. * = $p < 0.05$, ns = $p \geq 0.05$.

7.5 Peak Trunk Flexion

While ascending stairs, KECU and peak trunk flexion angles during the stance phase showed no correlation [$r = -0.06$; $p = 0.77$]. While descending stairs, KECU and peak trunk flexion angles during the stance phase suggests a weak negative correlation [$r = -0.31$], where those with a higher KECU, showed less peak trunk flexion, but was not supported statistically [$p = 0.14$].

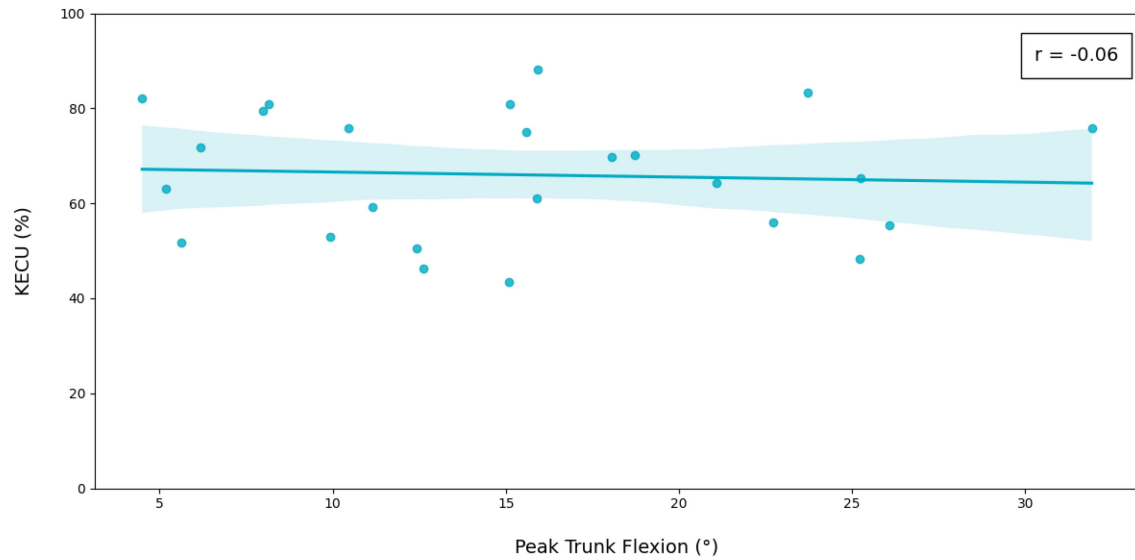


Figure 6.4: Lack of relationship between KECU and peak trunk flexion during the stance phase of stair ascent ($r = -0.06$; $p = 0.77$).

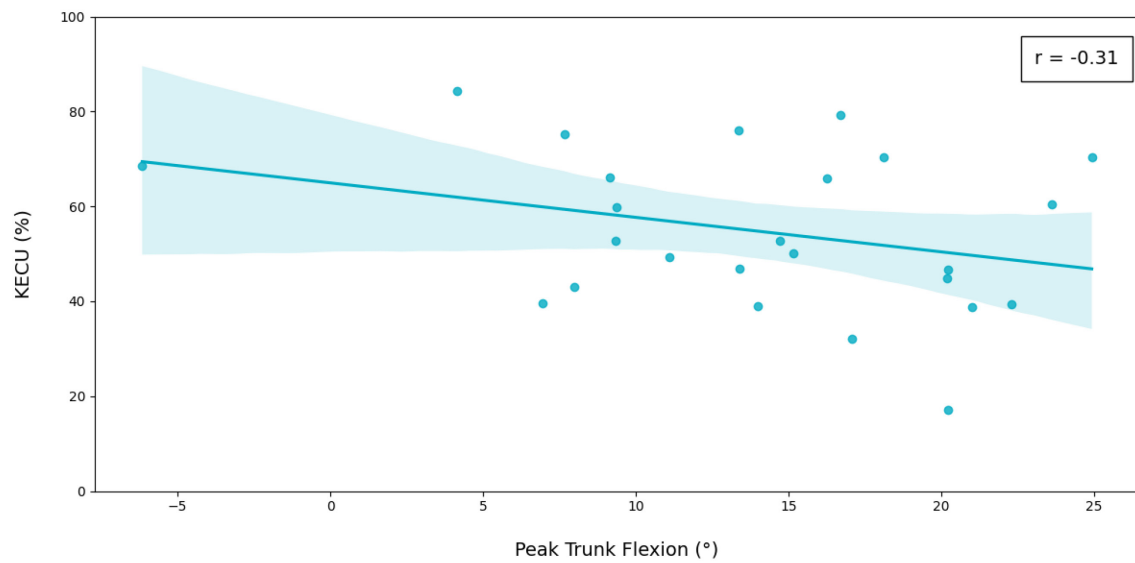


Figure 6.5: Lack of relationship between KECU and peak trunk flexion during the stance phase of stair descent ($r = -0.31$; $p = 0.14$).

8 Discussion

This thesis characterized KECU, along with the terms used to calculate KECU, during stair ascent and descent in healthy young and healthy older adults. To achieve this, I contributed to the design and build of a custom-built force plate instrumented 4-step staircase. Findings demonstrate that healthy older adults utilize a greater amount of their available knee extensor capacity than healthy younger adults during stair ascent but not during stair descent. This thesis also sought to discern if there is a relationship between KECU and sagittal plane trunk flexion. No significant association was found for stair ascent or descent. These findings may enhance our understanding of how individuals across different age groups utilize knee extensor capacity during dynamic tasks that impose greater demands than level walking. From a clinical perspective, this insight may be valuable for assessing and monitoring the extent of functional impairment associated with age-related declines in muscular function. However, the current evidence provides limited insight into potential trunk compensation strategies that may be employed to mitigate knee extensor weakness during stair negotiation.

While previous studies investigating knee extensor capacity utilization, also termed “functional demand” or “relative effort”, have focused on level-gait, the role of knee extensor muscle capacity utilization on stair-gait biomechanics remains unclear. This lack of clarity is primarily because of the wide range of estimates that previous research on knee extensor capacity utilization has demonstrated, resulting in low confidence when cross-referencing results from the current study with what is demonstrated by the literature. Knee extensor capacity utilization estimates for level-gait are between 20-66% and 10-94% for young and older adults, respectively (Hafer & Boyer, 2020; Homes & Boyer, 2022; Spinoso et al., 2019). Collectively, estimates for knee extensor capacity utilization during stair ascent are reported to be between 53-54% and 75-103% for young and older adults, respectively (Reeves et al., 2009; Hortobágyi et al., 2003). This study found KECU values of $62.1\% \pm 13.8\%$ and $73.0\% \pm 8.8\%$ for young and older adults, respectively. During stair descent, estimates are reported to be between 30-42% and 42-120% for young and older adults, respectively (Hortobágyi et al., 2003; Reeves et al., 2008; Samuel et al., 2011).

This study found KECU values of $57.2\% \pm 16.8\%$ and $57.6\% \pm 14.4\%$ for young and older adults, respectively. Interestingly, this thesis found both groups to exhibit lower knee extensor capacity utilization in stair descent than in stair ascent, which may suggest that ascending stairs is more demanding of the knee musculature than descending stairs. The lower estimates found in the literature agree with this thesis' findings, however the large range of estimates leaves uncertainty around the “true” normative estimates. Despite this variability across studies, this study and the existing literature consistently demonstrate that older adults utilize a greater proportion of their available muscle capacity during mobility tasks. The current study contributes novel data showing older adults use a greater KECU when ascending stairs.

The variability in KECU estimates found within the literature is likely a reflection of the variability in the methods chosen to measure KECU. For example, Reeves et al (2008; 2009) measured muscle capacity isokinetically at varying angular velocities and estimated knee extensor capacity utilization from approximating the angular velocity achieved during stair-gait to the nearest angular velocity isokinetically. Other studies have measured muscle capacity from MVICs at one or various knee flexion angles (Hortobágyi et al., 2003; Samuel et al., 2011). Notably, Samuel et al (2011) employed a linear interpolation method to enable a more precise estimation of knee extensor maximal capacity at the specific joint angle corresponding to the peak EKFM. Methods for measuring maximal capacity that fail to closely replicate the contraction dynamics of stair-gait, are likely to result in estimation bias of the muscle's maximal capacity. This, in turn, can lead to misinterpretation of muscle capacity utilization and the mechanical demands placed on the musculature while ascending or descending stairs. The methods used in this thesis were the first to estimate KECU by matching angular velocities at peak EKFM during stair-gait to a subject-specific third-order polynomial fit from both isometric and isokinetic data, which has been previously described for level-gait by Hafer & Boyer (2020).

After conducting this study, I recommend future work apply methods that best fit the chosen sample and task. The first consideration should be the demands of the task. More specifically, the

anticipated maximal joint angular velocities and the direction of these angular velocities that are to be expected for a given task should be clearly defined. Secondly, establish whether there are any potential limitations to the sample of interest that might prevent working at a specific angular velocity isokinetically. This will guide the decision to use a method which measures capacity isometrically or isokinetically. Ultimately, the chosen method of measuring capacity should match the task demands but only if the sample can reasonably complete the task designed to quantify maximal muscle capacity. This might lead to a decision to measure capacity using a method which does not accurately reflect the dynamic nature of a task (i.e., using isometric capacity values for a highly dynamic task), but fits the anticipated ability of the sample of interest to tolerate the procedures required. For example, if the joint demands of a task require high angular velocities, frail older adults may not be able to meet the demands of an isokinetic test at high speeds required due to functional or mobility limitations. In this circumstance, it may be prudent to measure isometric maximal capacity at the expense of using a metric that is less reflective of the task.

The methods outlined by Hafer & Boyer (2020), and within this thesis, are likely the most robust method for estimating maximal muscle capacity dynamically. A third-order polynomial, when fit to both eccentric and concentric capacities, reflects the relationship between joint moments and velocity. Given that muscle-tendon unit properties and neuromuscular control strategies are known to shift with age, this method may more accurately characterize adapted muscle dynamics across a spectrum of the lifespan. Additionally, the curvilinearity of a third-order polynomial offers a more physiologically realistic representation of the joint moment–velocity relationship compared to linear interpolation, which may oversimplify the nonlinear behavior observed across contraction types and velocities. However, linear interpolation may offer a better fit of the data when limited inputs (i.e., measured muscle capacities) are available (Hafer & Boyer, 2020). In cases where cubic interpolation proves unsuitable for estimating muscle capacity, as observed in a subset of participants in this study, the MVIC remains a likely valid modality for quantifying muscle capacity utilization. This is particularly relevant for tasks involving

eccentric muscle activation, such as knee extensor activation during descent, where MVICs tend to approximate peak eccentric knee joint moments more closely. Conversely, MVICs are likely to underestimate maximal muscle capacity, and ultimately muscle capacity utilization, in tasks involving concentric muscle activation, such as knee extensor activation during stair ascent. Regardless, MVICs as a measure of maximal capacity provides physiologically and clinically meaningful insights into muscle function while constituting a highly feasible method for estimating muscle capacity utilization across diverse populations and contexts. Future work could directly compare the impact of different strategies to estimate knee extensor capacity on KECU calculations.

In the literature, older adults or adults with musculoskeletal disorders are often excluded from research due to comorbidities or an inability to complete rigorous testing protocols. Using protocols that are both safe and operate within a wider functional spectrum removes barriers for entry in research participation which will in turn help further our understanding of muscle capacity utilization in functionally limited adults. For this reason, the ideal approach for measuring muscle capacity utilization will be described here. In this approach, kinematics and kinetics during a task are collected prior to muscle capacity. This data is to be subsequently processed to obtain both the peak joint moment and the angle or angular velocity at which the peak joint moment occurred. The researcher is now granted the ability to collect muscle capacity at the exact joint angle or angular velocity at which the peak joint moment was observed. This proposed approach for measuring muscle capacity utilization reduces participant burden by reducing the demands imposed by isometric and isokinetic maximal muscle testing at the expense of requiring an additional study visit. However, increased participant and task specificity comes at the cost of reduced comparability. Specifically, it limits the ability to make robust comparisons or generalizations of maximal joint capacity across participants, samples, and contexts. This limitation arises from greater inter-individual variability in the joint angles or angular velocities at which peak joint moments are observed.

Methodological challenges were encountered during stair descent. Different strategies were used when descending stairs which impacted on the timing of the peak EKFM as well as the angular velocity at which this occurred. More specifically, peak EKFM values obtained in the controlled lowering phase of the movement typically concurrently demonstrated high knee joint angular velocities that exceeded the knee maximal eccentric knee joint angular velocity tested via dynamometry (i.e., 180 deg/s). This required an alternative approach to determine KECU. This could have been mitigated by controlling stair-gait speed, ensuring knee joint angular velocities remained within the desired range. Alternatively, higher knee joint angular velocities could have been measured eccentrically on the dynamometer. Neither of these methods seemed viable due to the research questions posed and the population of interest. These methodological challenges made the data “noisier”; but nonetheless, this is one of the few known studies on KECU to provide data on stair descent.

Knee extensor capacity utilization could be a meaningful measure of muscle function that reflects the mechanical effort needed to accomplish a given task, such as activities of daily living. Clinically, this could be important for both monitoring the degree of functional impairment associated with age-related decreases to muscular function. It could also potentially be a risk factor for the incidence of knee OA. Initial observation of knee extensor weakness has been associated with the development of knee OA at follow-up within a 2.5-14-year period (Øiestad et al., 2015). For every 13.56 Nm increase in knee extension capacity, the odds of developing symptomatic knee OA may be lowered by 29%, by 20% for radiographic knee OA (Slemenda et al., 1997). In a 12-week strengthening program, females with symptomatic knee OA were found to increase knee extensor MVIC by 7.2 ± 15.0 Nm ($p = 0.02$) which was paired with a concomitant KECU decrease of $7\% \pm 13\%$ ($p = 0.008$) for walking and $0.61\% \pm 0.95\%$ ($p = 0.002$) for squatting tasks (Maciukiewicz et al., 2024). Indeed, KECU is likely elevated in people at risk for knee OA, relative to their peers. Holmes and Boyer (2022) found that participants with symptomatic knee OA experience higher functional demand when walking at a preferred gait-speed (36%) versus a faster gait-speed (50%). If the muscular demand on the knee extensors nears or meets its

available capacity, compensatory movement strategies may be necessary to successfully meet task demands. Thus, as task demands increase and/or muscle capacity decreases, there is a greater possibility that unfavorable movement strategies are adopted or the ability to react to neuromuscular stimuli is limited (i.e., a lower “reserve”). This may increase risk of musculoskeletal injuries, which may then hinder an individual’s ability to safely participate in physical activity; further contributing to degenerative joint diseases like knee OA.

With age-related musculoskeletal decline, adults may be more likely to utilize the trunk as a compensatory mechanism (McGibbon & Krebs, 2001). Increased trunk flexion over the ipsilateral or stance limb shortens the moment arm from the vertical GRF vector to the knee joint center, thereby reducing the EKFM and mechanical knee extensor demand (Iijima, Eguchi, et al., 2019; Iijima et al., 2018; Trinler et al., 2016). Altering trunk positioning during movement has been identified as a potential strategy to minimize knee joint loads (Hunt et al., 2008, 2010; Iijima, Shimoura, et al., 2019). Many studies have focused on the impact of trunk lateral flexion/lean on knee mechanics. For example, lateral trunk lean is associated with lower external knee adduction moments in people with medial knee OA (Hunt et al., 2008). Some studies have explored the associations of sagittal plane trunk kinematics on knee mechanics (Preece et al., 2019; Preece & Alghamdi, 2021; Turcot et al., 2013). One study of healthy females during stair ascent found that increased trunk flexion decreased patellofemoral joint stress rate and magnitude while increased trunk extension increased patellofemoral joint stress rate and magnitude (Atkins et al., 2019). Asay et al. 2009, found increased trunk flexion to be related to knee OA disease severity as those with greater severity also had reduced external knee flexion moments. During gait, forward trunk flexion is greater in knee OA compared to healthy controls (Preece et al., 2019; Preece & Alghamdi, 2021; Turcot et al., 2013). While it was anticipated that those working at a greater proportion of their available knee extensor capacity would demonstrate greater trunk flexion because of the role trunk flexion might have on lowering knee demands, no statistically supported relationship between KECU and peak trunk flexion during stair ascent or descent was observed. Since this study specifically

investigated peak trunk flexion and unilateral KECU, there are reasonable explanations as to why no observation was found.

First, it should be restated that the trunk was modelled as a single rigid body and the trunk angle reported in this study is measured from the global vertical axis. It is unclear if the trunk's relation to the pelvis, or the relation of any adjacent spinal segments to each other, may better reflect the movement of the trunk during this task. Additionally, as only the dominant leg during stance was assessed, it is unclear if the contralateral leg is influencing the demands placed on the study leg during stance. In stair ascent just prior to initial contact, the contralateral leg is helping support upward and forward acceleration of the COM in the forward continuance phase. In stair descent just prior to initial contact, the contralateral leg is helping support downward and forward deceleration of the COM in the controlled lowering phase. It is unclear the extent by which the contralateral leg in the end of double-leg support is influencing knee joints loads on the ipsilateral knee regardless of trunk positioning. Similarly, as maximal isometric and isokinetic muscle strength was only measured at the knee, it is unclear the role of ipsilateral ankle plantarflexor or hip extensor utilization may be influencing knee extensor utilization during single-leg stance phases. Furthermore, different knee joint angular velocity strategies were identified at the point of peak EKFM during stair descent. Altering gait-speed, and associated joint angular velocities, may also be a strategy to modify the KECU should an individual have the appropriate neuromuscular control to employ such a strategy with minimal risk of an adverse event. Ultimately, there are several reasonable variables yet to be thoroughly investigated that may have modified the peak EKFM, and thus the KECU, regardless of trunk positioning.

Despite insufficient evidence to support the observed weak negative relationship between peak trunk flexion and KECU was not purely due to random variance, describing the relevance of this finding within this context may be of value. The negative directionality in stair descent could reflect the possibility that increased trunk flexion might reduce knee extensor capacity utilization by reducing peak EKFM, which coincides with the literature. However, greater trunk flexion during stair descent might position the

center-of-pressure outside of an individual's base-of-support, increasing fall risk. This would be of particular concern in older adults who may not have the same level of confidence or efficiency with stairs and to an even greater extent while descending as the likelihood of a more severe injury outcome is higher.

9 Limitations and Strengths

This thesis is not without limitations. First, this thesis is underpowered by its sample size and as a result more robust analyses could not be accomplished. Should there have been a larger sample size, further investigation on any sex differences could have been made to collect more insight on the influence of sex on knee extensor capacity utilization. Increasing sample size would reduce the probability of a Type II error. The healthy young adult and healthy older adult group sample sizes are also non-equivalent in size ($n=16$ versus $n=9$, respectively). The smaller older adult group may not have had enough observations to detect if an effect exists, increasing the probability of a Type II error. Conversely, observed differences between the healthy young and healthy older adult groups could also be attributed to sample size. Second, for the purposes of feasibility, this study was conducted as a single-visit and did not grant the ability for participants to practice the dynamometer tasks. This lack of practice may have reduced the reliability of the maximal knee joint moments obtained. Examples of this threat to reliability include that three participants were not able to reach isokinetic angular velocities of 180 deg/s concentrically (achieved 178, 178, and 175 deg/s). Additionally, while a third order polynomial is a solid strategy that reflects the dynamic demands of most activities of daily living, a more accurate curve fit to the data would have been possible if more speeds were tested at smaller increments should more time have been allotted to dynamometry testing. Third, it is also important to note that the external knee flexion moment is a key variable for knee extensor capacity utilization, and it is unclear how opposing muscles contribute via co-activation. Greater co-activation of the knee flexors (i.e., hamstrings) would reduce the EKFM, thus reducing KECU. Without understanding how much opposing muscles are contributing to the net knee joint moment, it is difficult to conclude what the actual knee extensor usage is. Lastly, although having the ability to set the desired stair parameters on a custom staircase is an advantage, it limits the repeatability of this thesis as differences in step depth and height likely influences stair-gait mechanics.

There are several strengths of this thesis. To my knowledge, this thesis is the first of its kind to assess knee extensor capacity utilization during both stair ascent and descent using a method which uses a third order polynomial fit to muscular capacity values obtained via dynamometry. It is also the first attempt to determine if there is a relationship between knee extensor capacity utilization and trunk kinematics. Furthermore, it adds to existing research investigating the influence of the trunk in the sagittal plane during stair ascent and descent. This contribution is particularly important for the limited research that has investigated stair descent and highlights some of the challenges intrinsic to this area of research.

10 Conclusion

The aim of this study was to evaluate knee extensor utilization of younger and older adults during both stair ascent and stair descent on custom-built 4-step staircase. Findings demonstrated that older adults utilize a greater proportion of their knee extensor capacity during stair ascent but not for stair descent. These findings further support existing literature suggesting older adults have greater difficulty negotiating stairs. This is the first study to measure knee extensor capacity utilization during stair-gait using a method which estimates the maximal capacity from a third-order polynomial fit to both isometric and isokinetic maximal voluntary contractions both eccentrically and concentrically. Using this approach provides an estimate of muscle capacity utilization that better reflects the dynamic moment-velocity relationship of joints when undergoing more demanding tasks such as stair negotiation. These findings have clinical relevance, as they may help monitor the progression of functional decline and serve as potential indicators for the onset or development of musculoskeletal conditions associated with age-related reductions in muscle function.

Additionally, this study aimed to investigate whether knee extensor capacity utilization was related to peak sagittal plane trunk flexion angles. Findings demonstrated no relationship between peak trunk flexion angle and KECU in either stair ascent or descent. As sagittal plane trunk flexion did not increase with greater KECU, these findings do not support the proposed mechanism by which increased trunk flexion may be used to minimize EKFM and subsequently reduce KECU. While this study is the first to explore muscle capacity utilization of a lower extremity joint with the kinematics of the trunk, further research is necessary to better understand the role of the trunk in stair negotiation.

11 Future Work

Future work should investigate the psychological factors that contribute to one's ability to negotiate stairs and how this is impacted by the ageing process. This might include the degree to which fear, confidence, and cognitive ability contribute to movement efficiency. It is unclear the extent to which this impacts our neuromuscular control patterns during challenging tasks. On this note, as there is limited understanding of co-activation during stair negotiation, future work may deem necessary to have a more accurate representation of muscle activation, where electromyography may be useful. There may be other practical applications to which muscle utilization could be applied to other muscles. This could be a useful means of understanding fatigue and monitoring joint loading in different domains. For example, ergonomists could use muscle capacity utilization to measure total joint loads over various exposure periods in different environments or positions that could guide injury risk models. One study has begun to investigate muscle capacity utilization of the shoulder which could have interesting applications in the field of ergonomics (Murgia et al., 2018). However, there are several methods proposed in the literature to measure and calculate muscle capacity utilization and there is currently no consensus regarding which method is best. It would be sensible to first determine the reliability and accuracy of each existing method across diversified applications and should be a prominent focus for future work in this area. Establishing normative values for KECU will enable reasonable comparisons to be made across studies. Efforts should also be directed toward the development of computational models that unify the diverse methodologies currently used to estimate muscle capacity utilization. Such endeavors will ensure physiologically valid meta-analyses across studies and populations, while also potentially enabling integration into predictive musculoskeletal models.

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Appendix A: Get Active Questionnaire



Get Active Questionnaire

CANADIAN SOCIETY FOR EXERCISE PHYSIOLOGY –
PHYSICAL ACTIVITY TRAINING FOR HEALTH (CSEP-PATH®)

Physical activity improves your physical and mental health. Even small amounts of physical activity are good, and more is better.

For almost everyone, the benefits of physical activity far outweigh any risks. For some individuals, specific advice from a Qualified Exercise Professional (QEP – has post-secondary education in exercise sciences and an advanced certification in the area – see csep.ca/certifications) or health care provider is advisable. This questionnaire is intended for all ages – to help move you along the path to becoming more physically active.

- ☐ I am completing this questionnaire for myself.
- ☐ I am completing this questionnaire for my child/dependent as parent/guardian.

PREPARE TO BECOME MORE ACTIVE	
YES ↓ ○	NO ↓ ○
The following questions will help to ensure that you have a safe physical activity experience. Please answer YES or NO to each question <u>before</u> you become more physically active. If you are unsure about any question, answer YES .	
1 Have you experienced ANY of the following (A to F) within the past six months ?	
○	A A diagnosis of/treatment for heart disease or stroke, or pain/discomfort/pressure in your chest during activities of daily living or during physical activity?
○	B A diagnosis of/treatment for high blood pressure (BP), or a resting BP of 160/90 mmHg or higher?
○	C Dizziness or lightheadedness during physical activity?
○	D Shortness of breath at rest?
○	E Loss of consciousness/fainting for any reason?
○	F Concussion?
○	2 Do you currently have pain or swelling in any part of your body (such as from an injury, acute flare-up of arthritis, or back pain) that affects your ability to be physically active?
○	3 Has a health care provider told you that you should avoid or modify certain types of physical activity?
○	4 Do you have any other medical or physical condition (such as diabetes, cancer, osteoporosis, asthma, spinal cord injury) that may affect your ability to be physically active?
.....> NO to all questions: go to Page 2 – ASSESS YOUR CURRENT PHYSICAL ACTIVITY>	
YES to any question: go to Reference Document – ADVICE ON WHAT TO DO IF YOU HAVE A YES RESPONSE ...>>	

ASSESS YOUR CURRENT PHYSICAL ACTIVITY

Answer the following questions to assess how active you are now.

- 1 During a typical week, on how many days do you do moderate- to vigorous-intensity aerobic physical activity (such as brisk walking, cycling or jogging)? DAYS/
WEEK
 - 2 On days that you do at least moderate-intensity aerobic physical activity (e.g., brisk walking), for how many minutes do you do this activity? MINUTES/
DAY
- For adults, please multiply your average number of days/week by the average number of minutes/day: MINUTES/
WEEK

Canadian 24-Hour Movement Guidelines recommend that adults accumulate at least 150 minutes of moderate- to vigorous-intensity physical activity per week. For children and youth, at least 60 minutes daily is recommended. Strengthening muscles and bones at least two times per week for adults, and three times per week for children and youth, is also recommended (see csep.ca/guidelines).

GENERAL ADVICE FOR BECOMING MORE ACTIVE

Increase your physical activity gradually so that you have a positive experience. Build physical activities that you enjoy into your day (e.g., take a walk with a friend, ride your bike to school or work) and reduce your sedentary behaviour (e.g., prolonged sitting).

If you want to do **vigorous-intensity physical activity** (i.e., physical activity at an intensity that makes it hard to carry on a conversation), and you do not meet minimum physical activity recommendations noted above, consult a Qualified Exercise Professional (QEP) beforehand. This can help ensure that your physical activity is safe and suitable for your circumstances.

Physical activity is also an important part of a healthy pregnancy.

Delay becoming more active if you are not feeling well because of a temporary illness.

Appendix B: Descriptors & Anthropometrics

Age and Anthropometrics

Date of Birth (MM/YYYY):	__ / __
Body Mass (kg):	
Height (m):	
BMI (kg/m ²)	
*Waist Circumference (cm):	
Hip Circumference (cm):	
Resting Heart Rate (bpm):	
Dominant Leg: “Which leg would you typically use to kick a ball?” This knee will be designated “STUDY KNEE” if non-OA “healthy” group.	☐ Left ☐ Right
Involved Leg (OA ONLY): “Which knee gives you the most problems in terms of pain, stiffness, or difficulty completing activities during the day?” (Participant must select one) This knee will be designated the “STUDY KNEE” if OA group.	☐ Left ☐ Right

* Measure waist circumference at the point that is midway between iliac crest and lower rib margin, on maximal expiration.

Race

We aim to engage people of different races when studying health. Collecting this information will help us monitor and ensure diversity is represented in our studies.

We invite you to choose what best describes you (please **check all** that apply):

	Examples
☐ Black	African, Caribbean, Black Canadian, Afro-Latine, African American, or other African descent
☐ East Asian	Chinese, Korean, Japanese, or other East Asian descent
☐ Indigenous	First Nations, Métis, or Inuit
☐ Latine/Latinx	Latin American, Hispanic descent
☐ Middle Eastern	Afghan, Egyptian, Iranian, Lebanese, Turkish, Kurdish, or other Arab or Persian descent
☐ South Asian	East Indian, Pakistani, Bangladeshi, Sri Lankan, Indo-Caribbean, or other South Asian descent
☐ Southeast Asian	Filipino, Vietnamese, Cambodian, Thai, Malaysian, Indonesian, or other Southeast Asian descent
☐ White	British, German, Ukrainian, or other European descent
☐ Another race category	Not described above
☐ Do not know	
☐ Prefer not to answer	
☐ Prefer to self-describe	

If you chose “prefer to self-describe”, please indicate your answer below:

Sex and Gender

What is your sex assigned at birth?

- ☐ Prefer not to disclose
- ☐ Female
- ☐ Male
- ☐ Intersex
- ☐ Not listed

If you chose "not listed", please indicate your answer below:

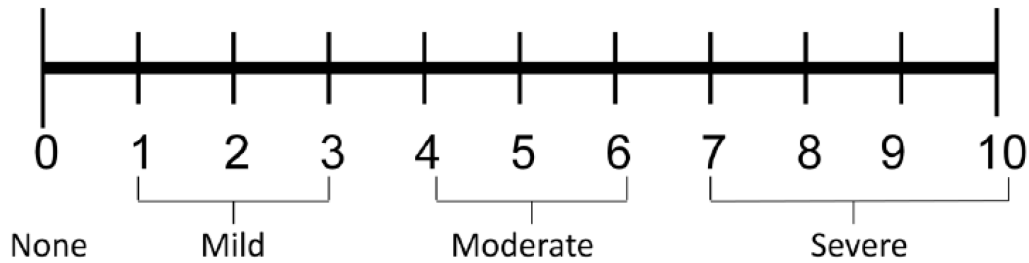
What is your current gender identity?

- ☐ Prefer not to disclose
- ☐ Woman
- ☐ Man
- ☐ Non-binary
- ☐ Prefer to self-describe

If you chose "prefer to self-describe", please indicate your answer below:

Appendix C: Numeric Pain Rating Scale (NPRS)

This survey will be administered to assess current pain upon arrival prior to any performance test (initial score) as well as after performance tests (final). Please indicate current pain intensity according to the scale below.



Initial Score (circle appropriate number)										
<u>Right Leg</u>										
0	1	2	3	4	5	6	7	8	9	10
<u>Left Leg</u>										
0	1	2	3	4	5	6	7	8	9	10
Final Score (circle appropriate number)										
<u>Right Leg</u>										
0	1	2	3	4	5	6	7	8	9	10
<u>Left Leg</u>										
0	1	2	3	4	5	6	7	8	9	10

Appendix D: painDETECT (PD-Q)

How would you assess your pain **now**, at this moment?

0	1	2	3	4	5	6	7	8	9	10
---	---	---	---	---	---	---	---	---	---	----

none

max.

How strong was the **strongest** pain during the past 4 weeks?

0	1	2	3	4	5	6	7	8	9	10
---	---	---	---	---	---	---	---	---	---	----

none

max.

How strong was the pain during the past 4 weeks **on average**?

0	1	2	3	4	5	6	7	8	9	10
---	---	---	---	---	---	---	---	---	---	----

none

max.

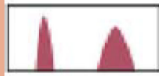
Mark the picture that best describes the course of your pain:



Persistent pain with slight fluctuations

☐


Persistent pain with pain attacks

☐


Pain attacks without pain between them

☐


Pain attacks with pain between them

☐

Please mark your **main area of pain**



Does your pain radiate to other regions of your body? yes ☐ no ☐

If yes, please draw the direction in which the pain radiates.

Do you suffer from a burning sensation (e.g., stinging nettles) in the marked areas?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

Do you have a tingling or prickling sensation in the area of your pain (like crawling ants or electrical tingling)?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

Is light touching (clothing, a blanket) in this area painful?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

Do you have sudden pain attacks in the area of your pain, like electric shocks?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

Is cold or heat (bath water) in this area occasionally painful?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

Do you suffer from a sensation of numbness in the areas that you marked?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

Does slight pressure in this area, e.g., with a finger, trigger pain?

never ☐ hardly noticed ☐ slightly ☐ moderately ☐ strongly ☐ very strongly ☐

(To be filled out by the physician)

never	hardly noticed	slightly	moderately	strongly	very strongly
☐ x 0 = 0	☐ x 1 =	☐ x 2 =	☐ x 3 =	☐ x 4 =	☐ x 5 =

Total score

out of 35

Please transfer the total score from the pain questionnaire:

Total score

Please add up the following numbers, depending on the marked pain behavior pattern and the pain radiation. Then total up the final score:



Persistent pain with slight fluctuations

0



Persistent pain with pain attacks

- 1

if marked, or



Pain attacks without pain between them

+ 1

if marked, or



Pain attacks with pain between them

+ 1

if marked



Radiating pains?

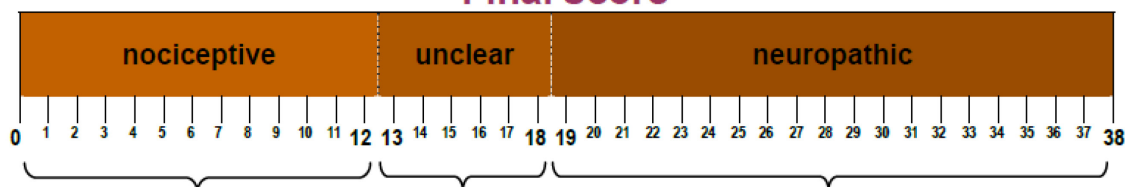
+ 2

if yes

Final score

Screening Result

Final score



A neuropathic pain component is unlikely (< 15%)

Result is ambiguous, however a neuropathic pain component can be present

A neuropathic pain component is likely (> 90%)

This sheet does not replace medical diagnostics.
It is used for screening the presence of a neuropathic pain component.

Development/Reference: R. Freynhagen, R. Baron, U. Gockel, T.R. Tölle / Curr Med Res Opin, Vol.22, No. 10 (2006)

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Appendix E: Knee Osteoarthritis and Outcome Scale (KOOS)

This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities.

Answer every question by ticking the appropriate box, only one box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms

These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never
☐

Rarely
☐

Sometimes
☐

Often
☐

Always
☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never
☐

Rarely
☐

Sometimes
☐

Often
☐

Always
☐

S3. Does your knee catch or hang up when moving?

Never
☐

Rarely
☐

Sometimes
☐

Often
☐

Always
☐

S4. Can you straighten your knee fully?

Always
☐

Often
☐

Sometimes
☐

Rarely
☐

Never
☐

S5. Can you bend your knee fully?

Always

☐

Often

☐

Sometimes

☐

Rarely

☐

Never

☐

Stiffness

The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None

☐

Mild

☐

Moderate

☐

Severe

☐

Extreme

☐

S7. How severe is your knee stiffness after sitting, lying, or resting **later in the day**?

None

☐

Mild

☐

Moderate

☐

Severe

☐

Extreme

☐

Pain

P1. How often do you experience knee pain?

Never

☐

Monthly

☐

Weekly

☐

Daily

☐

Always

☐

What amount of knee pain have you experienced the **last week** during the following activities?

P2. Twisting/pivoting on your knee

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P3. Straightening knee fully

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P4. Bending knee fully

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P5. Walking on flat surface

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P6. Going up or down stairs

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P7. At night while in bed

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P8. Sitting or lying

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

P9. Standing upright

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

A2. Ascending stairs

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

A3. Rising from sitting

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

A4. Standing

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

A5. Bending to floor/pick up an object

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A14. Sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc.)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A17. Light domestic duties (cooking, dusting, etc.)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

SP2. Running

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

SP3. Jumping

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

SP4. Twisting/pivoting on your injured knee

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

SP5. Kneeling

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never
☐

Monthly
☐

Weekly
☐

Daily
☐

Constantly
☐

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all

☐

Mildly

☐

Moderately

☐

Severely

☐

Totally

☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all

☐

Mildly

☐

Moderately

☐

Severely

☐

Extremely

☐

Q4. In general, how much difficulty do you have with your knee?

None

☐

Mild

☐

Moderate

☐

Severe

☐

Extreme

☐

Thank you very much for completing all the questions in this questionnaire.

Appendix F: Activities-specific Balance Confidence Scale (ABC-16)

Instructions: For each of the following, please indicate your level of confidence in doing the activity without losing your balance or becoming unsteady from choosing one of the percentage points on the scale from 0% to 100%. If you do not currently do the activity in question, try and imagine how confident you would be if you had to do the activity. If you normally use a walking aid to do the activity or hold onto someone, rate your confidence as it you were using these supports. If you have any questions about answering any of the following items, please ask the administrator.

0% 10% 20% 30% 40% 50% 60% 70% 80% 90% 100%

No Confidence Completely Confident

How confident are you that you will not lose your balance or become unsteady when you...

1. ...walk around the house?	%
2. ...walk up or down stairs?	%
3. ...bend over and pick up a slipper from the front of a closet floor?	%
4. ...reach for a small can off a shelf at eye level?	%
5. ...stand on your tip toes and reach for something above your head?	%
6. ...stand on a chair and reach for something?	%
7. ...sweep the floor?	%
8. ...walk outside the house to a car parked in the driveway?	%
9. ...get into or out of a car?	%
10. ...walk across a parking lot to the mall?	%
11. ...walk up or down a ramp?	%
12. ...walk in a crowded mall where people rapidly walk past you?	%
13. ...are bumped into by people as you walk through the mall?	%

Total Score:
(Sum %/ 16) %

Appendix G: 6-Minute Walk Test (6MWT)

BORG score BEFORE starting:

--

Instructions

“The object of this test is to walk as far as possible for 6 minutes. You will walk in a circle in this hallway for 6 minutes. Six minutes is a long time to walk, so you will be exerting yourself. You will probably get out of breath or become exhausted. You are permitted to slow down, to stop, and to rest as necessary. You may lean against the wall while resting, but resume walking as soon as you are able.”

“Remember that the object is to walk AS FAR AS POSSIBLE for 6 minutes, but don’t run or jog. Start now, or whenever you are ready.”

5 minutes remaining: “You are doing well. You have 5 minutes to go.”

4 minutes remaining: “Keep up the good work. You have 4 minutes to go.”

3 minutes remaining: “You are doing well. You are halfway done.”

2 minutes remaining: “Keep up the good work. You have only 2 minutes left.”

1 minute remaining: “You are doing well. You have only 1 minute to go.”

15 seconds remaining: “In a moment I’m going to tell you to stop. When I do, just stop right where you are.”

****If the patient stops walking during the test and needs a rest, do not stop the timer.**

Scoring

Distance in METERS covered in **6 minutes**:

--

BORG score AFTER walking:

--

Appendix H: 9-Step Stair Climb Test (9SCT)

Instruments

Standard 9 step stair case (with hand rail), digital stop watch, Numeric Pain Rating Scale.

Stair Ascent

Instructions: "Climb the stairs as quickly as possible, without compromising your safety. You may use the handrail. Do not run or jog. I will start timing as soon as your foot leaves the ground. I will stop timing as soon as both feet are planted on the top step. Ready...set...go."

- 1) Indicate the time needed to complete task is seconds

	Time (sec)	NPRS	Borg
Trial 1			
Trial 2			

- 2) Indicate the pattern of stepping with an 'X'

Alternating Steps	
Step Together	
Both alternating and step together	

- 3) Use of hand rail? YES ☐ NO ☐

Stair Descent

Instructions: "Descend the stairs as quickly as possible, without compromising your safety. You may use the hand rail. Do not run or jog. I will start timing as soon as your foot leaves the ground. I will stop timing as soon as both feet are planted on the ground. Ready...set...go."

- 1) Indicate the time needed to complete task is seconds

	Time (sec)	NPRS	Borg
Trial 1			
Trial 2			

- 2) Indicate the pattern of stepping with an 'X'

Alternating Steps	
Step Together	
Both alternating and step together	

- 3) Use of hand rail? YES ☐ NO ☐

Appendix I: Rating of Perceived Exertion (Borg CR-10 Scale)

Score	Level of exertion
0	No exertion at all
0.5	Very, very slight (just noticeable)
1	Very slight
2	Slight
3	Moderate
4	Somewhat severe
5	Severe
6	
7	Very severe
8	
9	Very, very severe (almost maximal)
10	Maximal

Appendix J: Stair Descent Knee Joint Dynamics Grouped by Stair-gait Cycle Phase

Table J1: Knee joint angular velocities at peak knee external flexion moment during descent grouped by the phase at which this occurs in the stair-gait cycle.

	Stair Descent	
	SLS (n = 15)	CL (n = 10)
Peak EKFM (N·m)	103.5 (29.0)	94.6 (31.6)
Joint Velocity @ EKFM (deg/s)	-83.3 (51.9)	-231.9 (31.5)

SLS = Single-leg Stance

CL = Controlled Lowering

EKFM = External Knee Flexion Moment