THE POWER OF EXERCISE: THE EFFECT OF AGE AND ACTIVITY LEVEL ON MUSCULAR POWER

DANIEL ALAIN DESROCHES

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Abstract

Rationale

Muscle, strength, and power decline as we age, and power is critical for functional independence. This dissertation tested additional factors such as the amount and type of PA, muscular fatigue, movement mechanics, and muscle fiber type, which are known to affect power, in older adults, including masters athletes.

Methods

One large collection, involving two lab visits, provided data for all research questions. Visit 1 (n=62, mean age 55.8±19.2y) consisted of PA questionnaires, endurance/control assessments followed by a biomechanical assessment of functional tasks including countermovement jumps (CMJ) with lower-body motion capture and a custom apparatus with embedded force plates. Visit 2 (n=36, mean age 57.8±17.6y) consisted of lower body MRI to measure muscle, fat, and diffusion, followed by a cardiorespiratory test.

Results

Age, sex and PA level predicted lower-body power during CMJ, with activity level demonstrating a protective effect (r=0.540) similar in magnitude to the effect of age (r=-0.654). Athletics discipline also predicted lower-body power during CMJ (r=0.389) with short distance athletes having the highest predicted power but also the most negative slope. No effects of age or activity level were found on change in lower-body power, which was used as a measure of muscular fatigue. Greater trunk flexion was associated with greater lower-body power, but older adults did not tend to use this strategy. MRI diffusion parameters weakly predicted ankle power and short distance athletes were shown to have different diffusion in the soleus and medial gastrocnemius when compared to non-athletes.

Discussion

Activity level and athletic discipline showed positive, protective effects on lower-body and joint power during the CMJ. The strength of association for activity level was comparable to that of age. Trunk flexion was associated with greater lower-body power output in the CMJ and was a strategy adopted only by younger adults potentially confounding the measurement of power in older adults. High levels of PA, and participation in high power track and field events is protective of muscular power and likely functional independence in older adults. Future fatigue investigations should consider the use of single joint power measurements alongside multi-joint movements such as the CMJ.

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List of Abbreviations

AHAbd	Active Hip Abduction
CI	Confidence Interval
CMJ	Countermovement Jump
CNS	Central Nervous System
COP	Centre of Pressure
CSA	Cross-Sectional Area
CT	Computed Tomography
DTI	Diffusion Tensor Image(ing)
DV	Dependent Variable
FA	Fractional Anisotropy
GRF	Ground Reaction Force
IC	Initial Contact
IQR	Interquartile Range
IV	Independent Variable
LTV	Lean Thigh Volume
MOS	Margin of Stability
MRI	Magnetic Resonance Image(ing)
MVPA	Moderate to Vigorous Physical Activity
PA	Physical Activity
PNS	Peripheral Nervous System
RD	Radial Diffusivity
RKP	Relative Knee Power
RMSD	Root Mean Squared Difference
ROM	Range of Motion
SLS	Single Leg Stance
TO	Toe Off

1 Introduction

The Canadian population aged 85 and older has doubled between 2001 and 2021. From 2016 to 2021, this population segment grew by 12% while the overall population grew by 5.2% (Statistics Canada, 2022). Increases in Canadian lifespan will increase demand for health care (Islam and Gilmour, 2024), and long-term care facilities (Statistics Canada, 2022) to accommodate the decreased physical ability associated with age. There are behavioral factors which can mitigate these burdens and, perhaps more importantly, improve the quality of life within this population segment.

Physical activity and/or exercise is one such behavioral factor. This dissertation explores how physical activity affects the expression of muscular power during functional tasks in a wide age-ranging cohort. Data was collected as one large study consisting of two visits. This dissertation has five distinct research questions and is written in manuscript-style chapters.

1.1 Rationale

Aging is accompanied by decreased muscle mass (1%/year), strength (2%/year), power (3.5%/year) [1] (Skelton et al., 1994), and the capacity to perform functional tasks (Hazell et al., 2007). These changes lead to increased frailty, loss of independence (Hunter et al., 2016), increased probability of falls (Katula et al., 2008; Simpkins and Yang, 2022) and decreased quality of life (Balachandran et al., 2022; Fragala et al., 2019). Perhaps most importantly, power is a superior predictor of function in older adults compared to strength or muscle mass (Balachandran et al., 2022; de Vos et al., 2005; Foldvari et al., 2000; Fragala et al., 2019; Skelton et al., 1994). As detailed in the next section, muscle power output and functional task performance are also affected by activity level, the type of physical training undertaken, one's movement strategy in demonstrating power, the extent of muscular fatigue present and muscle fiber type.

[1] The figure provided by Skelton and colleagues (1994) is likely exaggerated by the use of a higher reference age when calculating annual percentage changes. This effect is discussed as part of Research Question #1.

Less is known about *how* reductions in strength or power result in reduced functional task performance. To investigate this question, Hortobagyi and colleagues (2003) measured joint torques in 13 younger (~22y) and 14 older adults (~74y) as they performed functional tasks. The researchers demonstrated that when descending stairs, young adults required only 42% of their maximum knee extension moment compared to 88% in older adults. Although the authors did not measure joint power, their research suggested that joint-level analysis could guide task modifications and/or training programs for older adults. Secondly, it has been recently reported that within aging research, the true effect of aging is confounded by the effects of physical activity and/or sedentarism. To circumvent this issue, masters athletes have been proposed as a model for researching the true effects of aging, with less impact of physical inactivity (Lazarus and Harridge, 2017; Patelia et al., 2018). **Therefore, Research Question #1 evaluated peak specific lower body power, and joint powers during functional tasks and maximum countermovement jump (CMJ) in active and underactive adults.**

Recent emerging evidence suggests that power training (a form of resistance training) has beneficial effects on functional performance in older adults (Balachandran et al., 2022). Conveniently, masters athletes present a naturally divided cohort to test this idea in the wild, so-to-speak. Sprinters train for, and demonstrate bouts of brief, high-power outputs whereas endurance runners display longer lasting, lower power output (Rittweger et al., 2009). Existing research on power output in masters athletes shows no difference in peak power between masters endurance athletes and age-matched non-athletic controls (Piasecki et al., 2018). However, power-focused masters athletes have been shown to produce higher peak power compared to endurance masters athletes (Alvero-Cruz et al., 2021; Piasecki et al., 2018). **Accordingly, Research Question #2 determined the extent to which athletic discipline affected power output in masters athletes and non-athletes.**

It is important to consider the effects of fatigue on power output and functional task performance in older adults. Muscle fatigue is characterized as an exercise-induced reduction in maximal force or power which is specific to the task being performed (Enoka and Duchateau, 2008). A reduction in force or power is conceivably problematic in older adults, as additional reductions in physical capacity may further impair their ability to perform functional tasks. Paradoxically however, older adults are reportedly less fatigable than younger adults in *isometric* contractions of the elbow flexors and knee extensors (Christie et al., 2011). On the contrary, older adults are more fatigable than younger adults with *dynamic* contractions using decreased power as the index of fatigue. **Research Question #3 determined if activity level in older adults affects their fatigability as measured by decreased peak specific lower-body power during a CMJ.**

Movement strategies during a given task are also known to affect power output. Excessively 'vertical' trunk posture reduces countermovement jump height and results in more power being derived from the knees (Vanrenterghem et al., 2008). During gait, greater trunk flexion results in greater hip extension power (Teng and Powers, 2015). Similarly squat jump height has been reported to decrease by 14.4% on average when spine extension is removed from the forward dynamics simulation (Blanche and Monteil, 2013). To more completely understand the factors influencing power in this cohort, **Research Question #4 investigated the role of trunk and hip postures on power output during the CMJ.**

A final well-known factor that affects power output is that of muscle fiber type (Fitts et al., 1991). Individuals have varying proportions of the slow and fast twitch fibers which suggests that high-power performance can vary on this basis alone. For this reason, a novel MRI diffusion-based method was used to investigate the extent to which differences in power output may be driven by innate muscle physiology. **Therefore, Research Question #5 investigated MRI-diffusion parameters and their relationship to muscle power output.**

1.2 Research Questions and Hypotheses

Each of the five research questions introduced above have a set of hypotheses that were tested. These hypotheses are presented and tested in Chapters 5 through 9 and are revisited in the overall discussion, Chapter 10.

RQ #1: What is the effect of age, sex and activity level on lower-body specific power, and specific joint power during all functional tasks?

1.1. Specific lower-body power during the CMJ would decrease with age but increase with activity level.
1.2. Specific joint power during the CMJ and functional tasks would decrease with age but increase with activity level.

RQ #2: What is the effect of age, sex and athletics discipline on lower-body specific power and specific joint power during all tasks?

2.1 Lower-body specific power during the CMJ would decrease with age and would be highest in short distance athletes compared to all others.
2.2 Joint power during the CMJ would decrease with age and would be highest in short distance athletes compared to all others.

RQ #3: What is the combined effect of age, sex and activity level on changes in lower-body power output?

3.1 Underactive older adults (>50 years) will show fatigue in the post condition, quantified by reduced power in the CMJ.

RQ #4: What is the effect of hip and trunk posture on lower body power output?

4.1 Greater trunk flexion angle at peak GRF will be positively associated with lower-body specific power.
4.2 Greater hip flexion angle at peak GRF will be positively associated with lower-body specific power.

RQ #5: Can MRI-diffusion parameters predict muscular power output?

5.1 Fractional anisotropy (FA) of the soleus, lateral gastrocnemius and medial gastrocnemius muscle would be inversely proportional to ankle power output.
5.2 Short distance athletes will show lower FA in the plantarflexors compared to both non-athletes and long-distance athletes.
5.3 There will be a linearly decreasing correlation between the proportion of type I muscle fiber in each muscle and the measured FA for that muscle.

2 Literature Review

This review will first highlight some key physiological changes that typically occur with age. These include large and small-scale changes to muscle, decreased strength and power, and reduced functional ability. The potential to maintain or improve power and functional performance through various forms of physical activity (PA) is reviewed. In relation to functional performance, mechanics of common functional tasks such as the chair rise, stair ascent and stair descent are described. Surprisingly, there is limited data concerning joint power output during these tasks. Fatiguability has additional implications on power output and functional task performance in older adults. A brief review of the often-paradoxical literature related to fatigue in older adults is also provided for this reason. Lastly, discussion of reduced power in older adults would be incomplete without a review of potential mechanisms – the most notable being reduced muscle fibre size and altered fibre composition. This content also supports Research Question #5, which investigates a non-invasive MRI diffusion-based method that is reportedly capable of predicting muscle fibre type. Additional details surrounding masters athletes, and PA measurement methods are provided to support the preceding sections.

2.1 Quantitative Change to Muscle in Older Adults

Starting in the third decade of life, muscle mass decreases 3-8% per decade (Volpi et al., 2004) and can culminate in losses in males upwards of 50% by the 8-9th decade of life relative to 20-year-old male controls (Mitchell et al., 2012; Wilkinson et al., 2018). In healthy young adults, muscle mass is regulated via a dynamic balance between muscle protein synthesis (MPS) and muscle protein breakdown (MPB). Mechanical stimuli (exercise for example) and intake of protein rich foods stimulate MPS whereas the intake of carbohydrates reduces rates of MPB, allowing muscle mass to be maintained throughout the day (Wilkinson et al., 2018). Although nutrition plays an important role in the maintenance of muscle mass, the focus of this document will be around PA, and exercise more specifically, which provides the necessary mechanical stimuli for growth signalling. Regardless of age, limited movement, such as during bedrest or with sedentary behaviour, invites muscle atrophy via decreased MPS and/or increased MPB (Wilkinson et al., 2018). Low activity or inactivity likely affects muscle protein balance in older adults, as multiple studies have documented low participation rates of older adults in PA (McPhee et al., 2016), resistance training (Burton et al., 2017) and organized sports (Baker et al., 2010). In addition to the effects of inactivity on muscle mass, older adults are also impacted by anabolic resistance (Wilkinson et al., 2018). Anabolic resistance describes the blunted response to feeding, growth signalling and ultimately muscle growth (Kumar et al., 2009). Therefore, it appears likely that older adults lose muscle mass at a greater rate than younger adults at least in part due to reduced PA (McPhee et al., 2016) and can rebuild muscle less effectively due to anabolic resistance (Kumar et al., 2009; Wilkinson et al., 2018). In addition to decreased muscle mass, older adults also undergo qualitative changes to the whole motor unit (Hunter et al., 2016). These important changes have also been implicated in the impaired physical performance observed in older adults and are discussed next.

Key takeaway: Older adults aged 80-90 years will retain approximately 50% of their muscle mass. Losses of muscle mass are caused by several overlapping factors including, but not limited to, reduced participation in resistance training, PA and organized sports, changes in protein balance and anabolic resistance.

2.2 Qualitative Changes to Muscle in Older Adults

In addition to the quantitative changes to muscle presented above, older adults also undergo qualitative changes to muscle including preferential atrophy of Type II fibres (Lee et al., 2024), accrual of denervated angular fibres (Hunter et al., 2016; Roger Enoka, 1995), and infiltration of non-contractile tissue into and around the muscle cell (Overend et al., 1992; Perkisas et al., 2016; Rice et al., 1989). Most relevant to Research Question 5, this dissertation will only address changes to muscle fibre CSA and corresponding changes to phenotype proportion. It has been suggested that older adults show a reduced proportion of Type II (fast twitch) muscle fibres (Lexell et al., 1988). Some studies do show a decrease in Type II fiber proportion (Korhonen et al., 2006 [2]; Larsson et al., 1978; Nilwik et al., 2013), while others, including a 12-year longitudinal study, show a net gain in Type II fibre proportion (Frontera et al., 2000; Klitgaard et al., 1990 [3]). However, a majority of studies show no significant change and great variability in Type II fibre proportion with advancing age (Coggan et al., 1992; Frontera et al., 2008; Kirkeby, S and Garbarsch, C, 2000; Korhonen et al., 2006; Kraková et al., 2023; Messa et al., 2020). There are numerous factors (See **Table 2.1**) which confound these investigations such as the particular muscle being biopsied (Larsson et al., 2019; Naruse et al., 2023), biopsy site (Frontera et al., 2000), biopsy depth (Henriksson-Larsen et al., 1983; Korhonen et al., 2009), activity level (Lee et al., 2024; Purves-Smith et al., 2014), sex and nutrition status (Hunter et al., 2016), and measurement method (Andersen et al., 1999; Purves-Smith et al., 2014; Scott et al., 2001). A recent meta-analysis offers some clarity regarding fibre type changes with age and the major confounding factors of PA and sex. Lee and colleagues investigated myosin heavy chain (MyHC) protein expression, muscle fibre cross-sectional area (CSA) and fibre type proportion (%, based on fibre counts) in older versus younger adults across 27 studies and a sample size of 752 (Lee et al., 2024). When looking at MyHC protein expression proportions, the authors found an overall increase in MyHC I proportion and a decrease in MyHC II and IIA proportion in older adults compared to young adults. The authors also reported a non-significant change in Type I fibre CSA, and a decrease in both Type II and Type IIA fibre CSA between young and old participants. Categorizing participants as active or inactive demonstrated no differences in protein expression proportion between groups – both active and inactive older adults both showed increased MyHC I proportion, and decreased MyHC II and IIA proportion compared to active and inactive young adults. All fibre types showed reduced CSA in the active older adults but only Type II and IIA fibres showed reduced CSA in inactive older adults. Therefore, only inactive older adults showed preferential atrophy of Type II fibres. With respect to

[2] Korhonen et al. 2006 shows a decrease in Type II fibre proportion via histochemical analysis and no change via MyHC (myosin heavy chain) expression.

[3] Klitgaard et al. 1990 shows no change in Type II fibre proportion via histochemical analysis and decreased IIA in old runners and old swimmers via MyHC expression.

the effect of sex, the authors reported no significant differences in MyHC expression proportion between old and young females, whereas the male data was comparable to that of the complete dataset, with increased MyHC I and decreased MyHC II, and IIA when compared to young males. With respect to fibre CSA, the authors reported no significant sex differences – both older males and older females showed no significant change to Type I CSA, and a significant reduction in CSA of Type II and IIA fibers versus younger males and females, respectively. Fibre type distribution too was comparable to the complete dataset – both older males and older females showed non-significant changes in fibre proportion compared to young males and females respectively. Only five of 27 studies had female participants, therefore it is likely that more research is needed in females to understand sex differences. Of particular interest for this dissertation, several other studies have also demonstrated that Type II fibre CSA differences between young and healthy old participants can be nullified following 12 weeks (Kraková et al., 2023; Verdijk et al., 2016) to six months of resistance training (Nilwik et al., 2013). Measuring changes in fibre type with age is by no means simple but the best evidence suggests that older adults, especially those who are inactive, are likely to experience reduced CSA of Type II and IIA muscle fibres. These morphological changes are at least partially responsible for the changes in strength and power which are reviewed next.

Key Takeaways: Older adults show increased MyHC I, decreased MyHC II and decreased MyHC IIA protein expression in vastus lateralis. Active older adults experience atrophy of all major phenotypes whereas inactive adults exhibit preferential loss of Type II and IIA fibre CSA. Resistance training can reverse losses in Type II fibre CSA.

Table 2.1 – A summary of various resistance training (RT) interventions, cross-sectional studies and longitudinal studies related to change in muscle fibre type proportions, and fibre cross-sectional area (CSA) in young versus older adults.

Reference & *design*	**Sample**	**Changes per Histochemical Analysis**	**Changes per Myosin Isoform Content**
(Berg et al., 1997) *6-week bed rest intervention* *Vastus lateralis*	n=7m, 28 ± 2y	↓ Type I fibre CSA after 6 weeks ↔ Type IIA or IIB fibre CSA after 6 weeks ↔ Type I, IIA or IIB fibre type % after 6 weeks	↔ in MyHC I, IIA or IIB fibre type prop. after 6 weeks in VL
(Nilwik et al., 2013) *RT intervention* *Vastus lateralis*	n=25m, 23 ± 1y n=26m, 70 ± 1y	↓ Quadriceps muscle CSA in old vs young at initial (14%) ↓ Type II fibre CSA in old vs young at initial (29%) ↔ Type I fibre CSA following 6 months RT ↑ Type II fibre CSA following 6 months RT (24%)	n/a
(Verdijk et al., 2016) *RT intervention* *Vastus lateralis*	n=14m, 26 ± 2y n=16m, 72 ± 1y	↓ Type II CSA in old vs young at pre ↑ Type II CSA in old post vs old pre ↔ Type II fibre CSA in old vs young at post	n/a
(Kraková et al., 2023) *RT intervention* *Vastus lateralis*	n=20m, 21 ± 2y n=20m, 70 ± 4y (pre and post)	↔ in fibre type prop. b/w old and young in VL ↑ Type II CSA vs Type I CSA in young ↓ Type II CSA in old pre vs young ↓ Type II CSA vs Type I CSA in old pre ↑ Type II CSA in old post vs old pre	n/a
(Klitgaard et al., 1990) *Cross-sectional with MAs* *Vastus lateralis (& biceps br)*	n=7m young, 28 ± 0.1y n=8m old con, 68 ± 0.5y n=6m swim, 69 ± 1.9y n=5m run, 70 ± 0.7y n=7m weight, 68 ± 0.8y	↓ Type I prop in RT old vs all others ↑ Type IIA prop in RT old vs all others ↔ in prop between young vs old control	↑ MyHC I prop in old con vs young (27%) ↔ MyHC I prop in swim, run, and old con ↓ MyHC IIA prop in swim and run vs young
(Korhonen et al., 2006) *Cross-sectional with sprint* *Vastus lateralis*	n=16m sprint, 18-33y n=16m sprint, 40-84y	↔ Type I fiber CSA with age ↓ Type II fiber CSA with age ↔ Fibre type proportion with age	↑ MyHC I proportion with age ↓ MyHC IIx proportion with age ↓ MyHC I CSA with age (21%) ↓ MyHC IIA CSA with age (37%)
(Mosole et al., 2014) *Cross-sectional w mix MAs* *Vastus lateralis*	n=5m, young RT 26.2 ± 4.0y n=3m/3f, old sed 71.8 ± 3.5y n= 7m, mixed MA 68.3 ± 4.0y	n/a	↔ in fibre type prop b/w young RT vs old sed ↓ Type II prop and ↑ Type I prop in mixed MA vs others
(Cobley et al., 2016) *Cross-sectional w/ MAs* *Vastus lateralis*	n=6m young con (18-30) n=6m young cyclist (18-30y) n=6m old con (≥ 55y) n=6m old cyclist (≥ 55y)	n/a	↓ MyHC IIA proportion with age, regardless of training
(Messa et al., 2020) *Cross-sectional with mostly power MAs* *Vastus lateralis*	n=14m/8f, young con 19-27y n=27m/8f, old con 66-82y n=14m/0f, young ath 20-29y n=44m/7f, mid ath 38-65y n=32m/3f old ath 66-85y	↔ Fibre type prop. b/w old and young in VL ↔ Fibre type prop. b/w athletes and non athletes ↓ Type II fibre CSA with age in controls ↓ Type II fibre CSA with age in athletes (steeper slope)	n/a
(Frontera et al., 2000) *Longitudinal, 12y* *Vastus lateralis*	12m, 65 ± 4.2y at baseline 9m, 77.6 ± 4.0y at follow-up	↓ Type I fibre proportion after 12y (-20%) ↑ Type II fibre proportion after 12y (+20%) ↔ Type I or II fibre CSA	n/a
(Frontera et al., 2008) *Longitudinal, 9y* *Vastus lateralis*	12m/12f, 62-81y at baseline 5m/7f, 71-90y at follow-up	↔ Fibre type proportion ↔ Type I CSA ↔ Type IIA CSA	n/a
(Lexell et al., 1988) *Cross-sectional autopsy* *Vastus lateralis*	n=9m, 15-22y n=9m, 26-37y n=8m, 49-56y	↔ Type I fibre CSA with age ↓ Type II fibre CSA with age (26% from 20 to 80y) ↓ Total muscle CSA of 40% from 20 to 80y	n/a

	n=9m, 70-75y n=8m, 80-83y	↓ Muscle fibre count of 39% from 20 to 80y	
(Coggan et al., 1992) *Cross-sectional* *Gastrocnemius*	n=10m, 26 ± 1y n=10f, 23 ± 1 n=10m, 64 ± 1 n=10f, 63 ± 1	↔ in fibre type prop. b/w young and old ↓ Type IIA and IIA fibre CSA b/w young and old ↓ Type IIA and IIA fibre area ratios b/w young and old	
(Monemi et al., 1999) *Cross-sectional* *Masseter & biceps brachii*	n=5m, (19-25y) n=6m, (58-83y)	↓ Type I proportion (59 v 77%) in old adult masseter ↔ in fibre type prop. between old and young in biceps brachii	↓ MyHC I prop in old adult masseter (60 v 84%), but site specific ↑ MyHC I prop in old adult vs young biceps brachii (63 v 39%)
(Kirkeby, S and Garbarsch, C, 2000) *Cross-sectional, autopsy* *Masseter & vastus lateralis*	n=20m, 18-24y biopsy n=20m, 90-102y autopsy	↔ in fibre type prop. b/w old and young in VL and masseter ↓ Type I and II fibre CSA in old vs young VL ↓ Type II fibre CSA in old vs young masseter	n/a
RT: Resistance training(ed), **MA**: masters athlete, **con**: control, **ath**: athlete, **sed**: sedentary, **VL**: vastus lateralis, **prop**: distribution (%), ↑ greater/increased, ↓ lower/reduced, ↔ no diff/change Fibre type proportion (also referred to as distribution or composition) refers to a ratio based on fibre *count*. Fibre area ratio refers to a ratio based on fibre CSA.			

2.3 Changes to Muscular Strength in Older Adults

Beginning around 40-50 years of age, strength levels decline approximately 10% per decade (Hunter et al., 2016). Beyond age 65, losses can accelerate to 1-2% per year (Fragala et al., 2019; Skelton et al., 1994). As an example on the lower end, Hunter and colleagues (2016) suggest that an 80-year-old male would have approximately 40% of the strength of a 20-30 year-old male. On the higher end, Bassey and colleagues (1992) estimate that a male between the ages of 65-99 would retain approximately 50-60% of the handgrip strength compared to a young adult. Importantly though, there is some variability within these figures. For example, strength reductions were found to be greater in males, and greater in the lower limbs compared to the upper limbs (Goodpaster et al., 2006). Regarding the disparity in strength loss between lower and upper limbs, research has consistently demonstrated greater isometric strength losses in the knee extensors compared to the elbow flexors in older adults (Hunter et al., 2016) potentially due to differing usage rates in older adults between the two muscle groups.

Logically, strength levels also depend highly on PA and exercise status. One eight-year longitudinal study determined that regular, organized PA leads to significantly less leg strength loss in older females (age 65+) but this protective effect failed to reach significance in males (Gomez-Bruton et al., 2020). Of note, the PA reported in this study was defined loosely as a collective PA guided by an instructor and was possibly insufficiently rigorous to maintain strength in older males. In 2019, a large Canadian longitudinal study was published on active older adults (n=9100, age 60+) who regularly met Canada's PA guidelines for aerobic PA (≥ 150 min/wk). Within this large sample population, a subset of these adults who engaged in resistance training 1-7 days per week were shown to have greater strength (as measured by handheld dynamometer) than age-matched older adults who did no strength training, but still met Canada's activity guidelines (Copeland et al., 2019). It is relevant to note that handgrip strength has a strong association (R^2=0.74) with lower limb strength (Strandkvist et al., 2021) making it a relevant strength metric for many functional tasks.

Key Takeaway: Older adults (80+ years) retain approximately 50% of the strength relative to a young adult. Physical activity, especially resistance training, can reduce rates of strength loss in older adults.

2.4 Changes in Muscular Power in Older Adults

As mentioned in the previous section, older adults typically retain 50% of their muscle mass (Wilkinson et al., 2018) and older males typically retain 50% of their strength (Bassey et al., 1992; Hunter et al., 2016). Meanwhile, power retention ranges from 50% in athletes (Margaria et al., 1966), to 20-25% in long-term care home residents (Bassey et al., 1992; Bassey and Short, 1990). In 1966, Margaria and colleagues, investigated power output in an explosive sprint up a staircase. They recruited 131 male and female subjects aged 6 to 74 years, some of which were considered athletes being regularly active in sport. They did not report precise averages, but their graphical data demonstrate that athletes had consistently higher power output than non-athletes, that power output decreased steadily with age, and that there was limited difference in power output between males and females when controlling for bodyweight. They determined that power output appears to increase until 20-30 years of age with a power output of 15.7W/kg. Power output then steadily decreases to less than half of that by age 70 in both males and females (Margaria et al., 1966). In 1980, Bosco and Komi tested squat jumps, countermovement jumps, and depth jumps in a similarly wide-ranging population aged 4 to 73 years (n=226, 113m, 113f). During the squat jump, 18–28 year-old males and females produced 23.03 ± 4.20W/kg and 20.21 ± 2.99W/kg respectively, whereas 71–73-year-old males and females produced 6.98 ± 3.08W/kg and 4.98 ± 2.60W/kg. In the 1990's, a lab at the University of Nottingham tested the power output of a diverse-age population ranging from 20 to 93 years of age (Bassey and Short, 1990). In a subgroup of their sample aged 20-31 years (n=7, 4m + 3f) the peak power output as measured by their custom rig was 243 ± 67W. In a separate subgroup aged 74-91 years (n=32, 6m + 26f) the peak power output was 64 ± 42W. According to this data, the older group displayed approximately 25% of the power output of the younger population.

Key Takeaway: Lower body muscular power retention in older adults ranges from 50% in older athletes of approximately 70 years of age, to a mere 25% retention in 74–91-year-old non-athletes.

2.5 Power and Functional Performance

The previous section has demonstrated that muscular power decreases steadily with age in males, females, athletes and non-athletes alike. This section aims to describe the changes in functional performance that are theoretically tied to decreased power production in older adults. The goal of this section is to emphasize the importance of studying muscular power in older adults.

Muscular power is theorized to be more important for functional task performance compared to muscular strength (Balachandran et al., 2022; de Vos et al., 2005). Bassey and colleagues conducted one of the most cited studies on the topic to date (1992) where they measured power output and functional task performance in an older adult population (n=23, 80-99 years). They determined that the maximum single leg power output in males was 67.0 ± 8.3W compared to 34.8 ± 5.1W in females, or approximately 1.02W/kg and 0.64W/kg in males and females respectively. For comparison, the stair climb task required on average 80.4 ± 12.1W in males and 55.5 ± 10.1W in females. Based on these figures, it is no surprise that some participants were unable to complete the stair climb or chair rise (three of thirteen females, and one of thirteen men). Interestingly, Reid et al. performed a cross-sectional study comparing lower extremity power in healthy older adults (mean age 74 years) versus mobility-limited older adults (mean age 78 years). The mobility limited older adults demonstrated 65% less lower body power output (Reid et al., 2012). The findings from a 2016 systematic review suggest that lower limb muscle power explains a third to a half of the variance in functional performance in an older adult populations (Byrne et al., 2016). Muscular power is a strong predictor of functional capacity in older adults and as will be demonstrated next, it can also be improved, making it an important target for research.

Key Takeaway: Decreased muscular power production in older adults leads to decreased functional task performance and consequent limitations to mobility and independence.

2.6 Restoring Power and Functional Performance

The previous section has demonstrated that both muscular power and functional task performance decrease reliably with age. This section will demonstrate that various forms of PA are associated with the maintenance or improvement of power and functional capacity in older adults. Before proceeding, and for the purpose of this document, the following definitions are provided:

Physical activity (PA): “Any bodily movement produced by the contraction of skeletal muscle that increases energy expenditure above a basal level.” (Booth et al., 2012)

Metabolic equivalent of task (MET): The metabolic cost of resting quietly – One MET is approximately 3.5 mL of oxygen per kg body weight, per minute (Shephard, 2012).

Light-intensity physical activity (LPA): “…physical activity that is performed between 1.5 and 3.0 METs.” (Bull et al., 2020)

Moderate-intensity physical activity (MPA): “…physical activity that is performed between 3 and <6 times the intensity of rest (METs).” (Bull et al., 2020)

Vigorous-intensity physical activity (VPA): “…physical activity that is performed at 6.0 or more METs.” (Bull et al., 2020)

Exercise: “A subcategory of physical activity that is planned, structured, repetitive, and purposive in the sense that the improvement or maintenance of one or more components of physical fitness is the objective”. (Booth et al., 2012)

Sedentary Behaviour: “Any waking behaviour characterised by an energy expenditure of 1.5 METs or lower while sitting, reclining or lying.” (Bull et al., 2020)

In 2012, Ramsay and colleagues conducted a systematic review and meta-analysis on the effect of PA (including exercise) and sedentary behaviour (which differs from inactivity – see above) on various physical outcomes including lower-body power output. They found that step count, LPA, MVPA, and sedentary behaviour were positively associated with chair stand test performance (an indirect measure of lower-body muscle power) and direct measurement of lower-body muscular power via submaximal leg press or jumping. Direct measurement of lower-body power was also positively associated with total PA. Much of the older adult training research compares the effectiveness of traditional resistance training (TRT) and power training (PT), which involves medium loads, and a slow, controlled eccentric phase followed by an explosive concentric phase (Fragala et al., 2019). The literature suggests that both training modalities improve functional task performance in older adult, but PT does so to a greater extent. A systematic review by Byrne and colleagues (2016) analyzed the results of seven studies comparing PT with TRT for functional outcomes. They suggested that PT (i.e., performing resistance training exercises as fast as possible) will achieve a small to moderate advantage in functional task improvement when compared to TRT. A separate meta-analysis demonstrated that PT was more effective than TRT at improving performance in the chair rise and stair climbing tasks, but showed no benefit over TRT for walking speed and timed up and go performance (Steib et al., 2011). Another meta-analysis compared the evidence of PT vs TRT on functional performance in community dwelling older adults. The evidence suggested a small to medium effect on functional performance in favor of PT (Tschopp et al., 2011). Most recently, Balachandran et al (2022) carried out a systematic review and meta-analysis comparing the effects of PT and TRT on functional task performance in older adults. They examined whether PT was superior to TRT in improving physical function which was measured by a variety of physical tasks such as the 5x chair rise, 2.4m (8ft) get up and go, timed up and go, 30s chair stand, among others. Not all outcomes favored PT, but the overall standard mean difference favored PT. Of note, PT was also superior in improving self-reported functional performance (Balachandran et al., 2022).

Kew Takeaway: Power training (PT) can improve muscular power and functional task performance in older adults to a greater extent than traditional resistance training (TRT).

2.7 Masters Athletes

Masters athletes are a unique and convenient cohort of highly active older adults, who have in some cases been highly physically active for their entire adult lives. For this reason, they are thought to display the effects of age on various physical capacities without the effects of inactivity as a confounding factor. Within the world of Masters track and field, there exist various competitive events, categorized as long, middle and short distance, which are convenient gradations of power output. This convenient detail allows for the study of the effects of PA but also the *type* of PA on muscular power output. These track and field categories are used for this purpose in Research Question #2 and #5. Several of the most relevant events and their categorizations are shown below in **Table 2.2**.

Table 2.2 - Categories of track and field disciplines per World Athletics.

Long	Middle	Short
Running 3000m+ Race walking 3000m+ Triathlon	Running 800 to 3000m Relay 4x800m	Sprinting (60, 100, 200, 400m) Jumping (triple, high, long, pole vault) Throwing (shotput, discus, javelin, hammer, weight) Combined events (decathlon, heptathlon, pentathlon) Relay 4x100, 200, 400m
*Combined events typically contain one middle distance run event, but a majority of short-duration events.		

2.8 Measuring Physical Activity

In general, PA can be measured in two ways – direct (or objective), and indirect (or subjective). Direct monitoring can be achieved by physiological measurement such as heart rate or VO_2 assessment, or by the measurement of movement, with the most common method being accelerometry (known as inertial measurement units if additional sensors are used; Kowalski et al., 2012). Accelerometers quantify PA by detecting the frequency and magnitude of acceleration(s) of a body segment within a specific window of time (epoch length). Following data processing which involves a minimum of on-board filtering and the use of algorithms, these devices can classify activity level intensity and duration (Migueles et al., 2017). As for downsides, these devices (and accompanying software) have a financial cost associated, can be intrusive to the participant, do not describe the type of activity, and are time-consuming to both participant and researcher. While they do not suffer from the same biases involved in surveys, accelerometers can cause individuals to alter their PA behaviours while wearing the monitor (Kowalski et al., 2012). Perhaps the most significant concern however is that accelerometers are not suitable for measuring non-ambulatory activities such as resistance training and cycling (Cleland et al., 2018). Lastly, data processing decisions need to be made regarding selection of filters, epoch length, non-wear-time definition, cut-points, and algorithms. Algorithms are usually selected based on a narrow age group (i.e., algorithms trained for children versus older adults). This is a particularly challenging issue for this study's cohort due to the wide age-range. As for the benefits, accelerometers are almost completely objective (as above, researchers must choose algorithms) and free of recall bias (inaccurate recall of daily or weekly PA) or social desirability biases (Brenner and DeLamater, 2014), a form of response bias which involves responding to surveys in a way that is to be viewed positively by society (Teh et al., 2023). Objectivity of measurement is critical as self-reported measures have been shown to have unsystematic error, both under- and over-estimating PA in various studies, making correction difficult (Prince et al., 2008).

Subjective measures of PA refer to surveys or questionnaires. In addition to being less invasive and more cost-effective than accelerometers or other objective measurement methods, surveys such as the IPAQ are also well-accepted and include non-ambulatory activities such as weight lifting and cycling. A 2011 systematic review assessed the validity of the IPAQ-SF by comparing walking, moderate and vigorous intensity minutes measured by accelerometer or pedometer to that of the IPAQ survey. They found generally small correlations, with three instances of vigorous minutes, one instance of moderate minutes, and two instances of walking minutes achieving a correlation greater than 0.40 against the objective measure (out of 74 possible correlations). IPAQ-SF vigorous minutes achieved a correlation of 0.41 to $VO_{2,max}$ (Lee et al., 2011) although in a separate study, Fogelholm and colleagues described a paradoxical finding where participants in the highest PA quintile had a $VO_{2,max}$ matched to that of the second quintile (2006). The authors indicated that a significant proportion of the most active individuals had over-reported vigorous PA. This likely contributes to the unsystematic error reported by Prince and colleagues (2008), discussed previously.

The HUNT survey, originally part of a Norwegian population study, is a simple three-question exercise survey which has been shown to have a correlation of 0.48 with VO_{2max}, 0.39 to vigorous METs (measured by an ActiReg accelerometer), 0.55 with IPAQ vigorous minutes and 0.30 to IPAQ moderate minutes in a sample of 108 males aged 20 to 39 years (Kurtze et al., 2008). A subtle benefit of the HUNT survey is its inclusion of perceived effort to rate one's intensity during exercise in lieu of the type of activity dictating the intensity. Kowalski and colleagues (2012) stressed that perceived intensity will differ depending on an individual's age and fitness level. Therefore, a survey using perceived effort may be most appropriate for a cohort with a wide age and fitness range.

Key Takeaway: Physical activity can be measured directly (most commonly with accelerometers) or indirectly via questionnaire. Accelerometers are almost entirely objective, eliminating many biases present in surveys, but are invasive and do not adequately capture activities such as resistance training and cycling. Questionnaires are less invasive but are subject to several response biases.

2.9 Biomechanics of Functional Tasks

An overview of existing biomechanical analyses of functional tasks may help clarify the relationship between power output and functional task performance in older adults. These analyses have been conducted for more than six decades, but joint power has received little attention. The most heavily analyzed tasks are the sit-to-stand (or chair rise), stair ascent and stair descent. All three of these tasks have mechanical differences that make them challenging in unique ways. For example, the sit-to-stand task relies predominantly on hip joint moment regardless of seat height (Rodosky et al., 1989), stair climbing relies predominantly on knee moment (Nadeau et al., 2003), and stair descent requires less knee moment but greater knee ROM compared to stair ascent (Hortobagyi et al., 2003). Time spent in single support during stair ambulation surely also presents a unique challenge to the older adult. A more detailed review of the mechanics involved in these tasks is presented here, with a focus on their performance by an older adult population. A tabular summary of the main numerical findings for these functional tasks is provided in **Appendix A**.

2.9.1 Sit to Stand Task

Rising from a chair is considered one of the most important requirements of independent living (Ellis et al., 1984; Rodosky et al., 1989). There have been several biomechanical analyses over the past many decades looking at a combination of joint motion, joint moments, movement strategies, relative effort, and even chair design. In 1994, Carr and Gentile investigated various arm constraints while rising from a chair in young adults (24 ± 2.3 years). In the preferred condition, where participants could use their arms freely, they demonstrated that joint power consistently peaked in the order of hip, knee, and ankle. They also demonstrated that this ordering of peak power was disrupted when the arms could not be used (Carr and Gentile, 1994). Rodosky and colleagues (1989) used an instrumented chair, force plates and motion capture to measure lower body joint motion and joint moments in ten healthy adults aged 25.5 years on average. The highest moment occurred at the hip regardless of chair height, and as chair height decreased, they reported increasing knee moments. With the seat pan adjusted to 65% of knee height versus 115% of knee height, they observed a doubling in knee moment (Rodosky et al., 1989). In 2003, Hortobagyi and colleagues assessed relative effort in several activities of daily living including the chair rise (Hortobagyi et al., 2003). Relative effort was measured as joint moment during chair rise divided by the joint moment obtained during a maximum effort isometric leg press done at a similar joint angle. On average, older participants required 80% of their maximal knee extensor torque capacity to rise from a chair compared to 42% in younger participants. In 2016, Glenn and colleagues conducted a functional power assessment in the chair rise task in 13 older sedentary individuals (59.3 ± 4.5 years), 35 older recreationally active individuals (59.6 ± 5.0 years) and 26 masters athletes (56.7 ±

5.4 years). Participants were instructed to rise from the chair as quickly as possible. There were no significant differences in average power or average velocity however peak power and peak velocity were significantly greater in masters athletes compared to both sedentary and recreationally active participants (Glenn et al., 2016). Of all the studies mentioned above, only Carr and Gentile (1994) measured joint powers, albeit only six subjects. Given their intriguing findings with respect to the ordering of joint powers, and the importance role of muscular power output in functional task performance, it appears fruitful to build upon their analysis with a larger sample size, and modern methodology.

Key Takeaway: Hip moments dominate when rising from a chair, but knee moments increase with decreasing seat height. Joint powers peak in the order of hip, knee, and ankle when arms are used freely. Masters athletes demonstrate higher peak power when rising from a chair as fast as possible compared to sedentary and active older adults.

2.9.2 Stair Ascent and Descent

Stair climbing is a form of locomotion characterized by large sagittal plane moments and powers which drive the body forward but also upwards against gravity (Nadeau et al., 2003). For older or otherwise less capable individuals, stair ascent and descent can pose a serious challenge. A review of the literature related to these challenging mechanical requirements for stair climbing is presented here.

Hortobagyi and colleagues evaluated sagittal plane knee moment in older adults during functional tasks including stair ascent and descent (Hortobagyi et al., 2003). They determined that on average, the older adult participants were operating near their maximum capacity for knee moment while ascending and descending stairs, requiring 78 to 88% of their maximum capacity, compared to 54 and 42% for young adults. A separate study investigated hip torque in older adults (n=30, 65.4 ± 6.02 years) during walking, stair ascent, descent, and various other tasks in all three planes of motion (Kirkwood et al., 1999). Sagittal plane hip extension torque during stair ascent and descent were comparable to that observed during level walking (0.82 vs. 0.77 N-m/kg) for walking vs stair ascent and 0.91 vs 0.96 N-m/kg, for walking vs descent). The first study to include joint power analysis of stair navigation was published by McFayden and Winter in 1988. They concluded that when ascending and descending stairs, hip joint moments and powers were dissimilar between subjects whereas ankle and knee moments and powers were largely stereotypic. This analysis was conducted on a sample size of three, so the results are not likely generalizable. Several years later, Nadeau and colleagues conducted a similar study, also in older adults (n=11, 41-70 years). They analyzed joint moments and powers at the ankle, knee and hip joints during stair ascent and level walking (Nadeau et al., 2003). They found that hip extensor moments were similar during walking compared to stair ascent whereas knee extensor moments were twice as high in the stair climb. They also reported greater hip and knee extensor power during stair climbing compared to level walking (Nadeau et al., 2003). Given the importance of muscular power on functional task performance, it stands to reason that research is needed in the realm of joint power during stair ascent and descent. As with sit-to-stand literature, future research on stair navigation should include a larger sample size, consisting of both males and females and more modern methodology.

Key Takeaway: Stair descent requires 88% of an older adult's maximal knee extensor moment. Hip extension moment in stair climb compares to level walking whereas knee extensor moments are twice as high in the stair climb. Hip joint moments and powers can vary between participants, but generally, greater knee and hip extensor power is required in stair climbing compared to walking.

2.10 Fatigue During Functional Tasks

Fatigue can be defined as decreased maximal force or power, whether or not the task can be sustained (Enoka and Duchateau, 2008; Vøllestad, 1997). It can involve both central factors which exert their effects proximal to the neuromuscular junction (NMJ), peripheral factors which exert their effects distal to the NMJ, or a combination of both (Aquino et al., 2022). Central factors can include neurochemical changes and reduced intrinsic motivation. A lack of intrinsic motivation can be measured and mitigated against to an extent by interpolated twitch techniques however this method requires stimulation at the nerve trunk or the muscle belly (Shield and Zhou, 2004) which makes its use in functional, multi-joint tasks difficult. It has been suggested that intrinsic motivation is a larger concern for longer tests or for tests of unknown duration as participants will develop a pacing strategy based on sensory feedback and the goals of the activity. These strategies impair maximum effort performances and thus, fatigue (Aquino et al., 2022). Peripheral factors include the accumulation of metabolites such as hydrogen ions, lactate and inorganic phosphates, energy deficits in the form of ATP and glycogen, and reduced oxygen concentrations (Aquino et al., 2022). Ultimately, all these factors work together to impair muscle contraction which can be measured as decreased maximal force, power, or both.

Considering that power is critical for functional performance, and is already reduced in older adults, it is easy to see how fatigue can be a concern for older adults. As a tangible example, older adults aged 80-99 years have been shown to produce 0.88W/kg on average while climbing stairs. It has also been suggested that there is a threshold power output of 0.5W/kg below which normal tasks are no longer feasible (Bassey et al., 1992). Therefore, although older adults may be able to initially complete a task such as stair climbing, fatigue may eventually compromise this ability.

Many researchers have measured fatigue in older adults and the results are complex due to the many different muscle groups, types of contractions, intensity of contraction, duty cycle, method of measuring fatigue, sex, muscle group and the use of electrical stimulation (Christie et al., 2011; Enoka and Duchateau, 2008). In their systematic review, Christie and colleagues (2011) determined that older adults are more fatigue resistant than younger counterparts during isometric contractions of the elbow flexors and knee extensors using endurance time as the index of fatigue. However, older adults appear more fatigable when the fatiguing protocol involves dynamic contractions, and when decreased power is the index of fatigue. Subsequently, Kruger and colleagues (2018) published an additional meta-analysis on the topic of fatigue in older versus younger adults and their results agreed with Christie and colleagues. They added however that the fatiguing protocol could also be an isometric task, but the index of fatigue must be power measured during a dynamic task. Most recently, Paris and colleagues (2022) conducted the third meta-analysis on the topic to date. They added that when using a dynamic fatiguing task as both the mechanism and index of fatigue (as opposed to the above-mentioned isometric mechanism, with a dynamic index), the dynamic task should be isotonic, and not isokinetic. They theorized that controlling angular velocity could limit the development of fatigue. Fatigue is an important concept in understanding the limitations in functional performance of older adults. However, as above, it must be measured carefully with decreased power as the index of fatigue, and with an isotonic task if a dynamic mechanism is used.

Key Takeaway: Using decreased power output as the index of fatigue is a useful discriminative method to capture fatigue in older adults.

2.11 Measuring Changes to Muscle and Fat with Magnetic Resonance Imaging

The preceding sections have demonstrated that muscular power is critical for functional performance in older adults. One distinction that must be made when measuring power is that of normalizing, or controlling for body mass, i.e., specific power. Specific power, typically normalized to body mass, is used extensively in research as it facilitates comparison between participants of differing body mass, and across cohorts (See **Appendix A**). One concern with power normalized to body mass is that an individual with high amounts of body fat, which includes a majority of older adults, would have their power underrepresented (Martin et al., 2000). Existing data suggests that the thigh and spine muscles atrophy and become infiltrated with fat as one ages (Engelke et al., 2022), and that intramuscular adipose tissue (intraMAT) is greater in age- and BMI-matched frail older adults versus non-frail (Addison et al., 2014). Failure to account for increases in adipose tissue can lead to overestimation of muscle CSA by as much as 12% (Elder et al., 2004). As well, previous research has shown that life-long male athletes have less muscle fat infiltration and greater muscle cross-sectional area than age-matched controls. Females had similar muscle fat infiltration compared to controls in all muscles except the pectoralis (Emanuelsson et al., 2022). In summary, greater fat infiltration has been observed in frail older adults, and lower fat infiltration has been observed in older male athletes. These trends would impair the interpretation of specific power in these two groups, and this cohort. One possible mediation for this is to calculate contractile tissue cross-sectional area (CSA) from MRI Dixon (imaging protocol which can separate muscle and fat) and use this instead of body mass to normalize power. In 2012, Reid and colleagues investigated several possible reasons for loss of muscular power in mobility limited older adults including measuring muscle CSA and fat infiltration via CT imaging. They found that healthy older adults (approximately 74 years) and mobility limited older adults (approximately 78 years) both showed reduced peak leg extensor power compared to healthy middle-aged participants (approximately 47 years). Mobility limited older adults showed lower muscle CSA, greater intermuscular adipose tissue (interMAT) and reduced specific power (W/cm^2) when compared to both healthy groups (Reid et al., 2012). Reduced specific power suggests that the impairments of muscular power were due to factors other than muscle atrophy. Similar muscle and fat CSA measurements can also be done via MRI, which does not expose participants to radiation. In 2017, Davison and colleagues investigated muscle fat infiltration in individuals with osteoarthritis using an MRI method to separate fat and water signals (GE IDEAL). This method allows for the separate quantification of subcutaneous adipose tissue, interMAT, and intraMAT within each muscle. Given the increased adipose tissue deposition in older adults, and the importance placed on specific power in research, it is likely beneficial to take a critical look at normalizing methods. The additional findings of Research Question #1 present this data.

Key Takeaway: MRI can accurately measure muscle and adipose tissue cross-sectional areas (CSAs). Moments and powers to be expressed relative to muscle CSA which allows comparison across participants and studies.

2.12 MRI Diffusion Parameters and Power Output

It known that high power athletes have a greater proportion of fast-twitch Type 2 fibers compared to elite endurance athletes who have a greater proportion of slow-twitch Type 1 fibers (Wilson et al., 2012). Type II muscle fibers produce higher power outputs (Fitts et al., 1991) but are also more sensitive to fatigue (Beelen and Sargeant, 1991). Muscle fiber type proportion may therefore help explain differences in performance independent of age and activity status in this cohort. A non-invasive methodology was selected as biopsies are known to affect subsequent physical performance (Lievens et al., 2020) and the discomfort involved would likely affect recruitment and drop-out rates in an older adult population.

MRI diffusion tensor imaging (DTI) allows for the assessment of molecular movement in a biological tissue. It is commonly used to analyze the microstructure of biological tissues such as fiber tracts in the brain, but it has also been used to non-invasively approximate the proportion of Type 1 and Type 2 fibers in human muscle (Scheel et al., 2013). The authors proposed that differences in microstructure and fiber size between Type 1 and Type 2 fibers are expected to contribute to the differences in the various DTI parameters, fractional anisotropy (FA), mean diffusivity (MD), parallel diffusivity (PD) and radial diffusivity (RD). Scheel and colleagues (2013) compared muscle tissue biopsy results from 12 healthy male participants with MRI DTI parameters of the same muscles. Fractional anisotropy (FA) and radial diffusivity (RD) were the best predictors of Type 1 fiber type proportion, having a correlation of 0.71 ($p<0.05$) and 0.72 ($p<0.001$). Based on their findings, they suggested that Type 1 fibers presented a greater hindrance to diffusion than Type 2 fibers in the radial direction due to of a greater amount of microstructure such as mitochondria. They also reasoned that type II fibers would allow for greater radial diffusivity owing to their larger average diameter. Following this work, the same group assessed the relationship between the same MRI DTI parameters against muscle power of plantarflexors in a position of knee flexion to isolate the soleus muscle (Scheel et al., 2013a). They determined that fractional anisotropy (FA) was the best predictor of power with a correlation of -0.85 ($p<0.01$). From their previously mentioned work, a higher FA value implied a greater proportion of Type I fibers, which from this work correlated with lower power output. In 2022, Takao and colleagues sought to investigate the effect of ankle position (dorsiflexed, neutral and plantarflexed) and contraction mode (non-contracted, neutral isometric, shortened isometric) on DTI measures of the soleus, in comparison to soleus power output measured via Biodex. They found that in general diffusion was greater in the non-contracted state however no significant relationships were observed between FA and power output in any ankle position or contraction mode (Takao et al., 2022). More recently, Cameron and colleagues (2023) assessed the relationship between MRI diffusion parameters and fiber typing using a different approach. They related FA in various thigh muscles to Type I muscle fiber proportion in the same muscles taken from existing literature. They showed that muscles with a lower proportion of Type I fibers such as semitendinosus and rectus femoris had greater FA and muscles with a greater proportion of Type I fibers such as the vastus medialis and intermedius showed lower FA. Interestingly, this is the opposite trend to be expected based on the earlier work of Scheel and colleagues. This inconsistency will be explored further in the discussion of RQ5.

Despite some inconsistencies, this novel, developing method may help to approximate muscle fiber composition which in turn may help to explain some differences in power output within this cohort.

Key Takeaway: MRI DTI can non-invasively approximate the proportion of Type 1 and Type II fiber types and is highly correlated (r= -0.85) to power output. This research is still developing but may help explain differences in maximal power and fatigue.

3 Common Methodology

3.1 Introduction

A description of the methodology shared by most research questions is provided here. Research questions which deviate from this common methodology will have additional pertinent information within their respective chapters. Research Questions #1, 3 and 4 rely predominantly on the common methodology presented here. Research Question #2 has a smaller sample size resulting from the exclusion of “other” athletes and Research Question #5 will have additional methods related to MRI and has a smaller sample size due to MRI disqualification as is discussed in the Inclusion Criteria section.

3.2 Methods Overview

Participants visited the lab twice roughly one week apart. The first visit involved anthropometric measures, core endurance testing, power testing, a functional task assessment and several surveys regarding PA, general well-being, PA history, and injury and/or surgery history. During the second visit, participants underwent MR imaging followed by a treadmill-based VO_{2max} test. In between visits 1 and 2, participants were asked to wear an Actigraph (GT9X) activity monitor 24h per day, for seven days. A subsample of the ten first participants were invited to return after six months time to repeat the first visit, to observe whether changes occurred over time. A general overview is provided below in **Figure 3.1**.

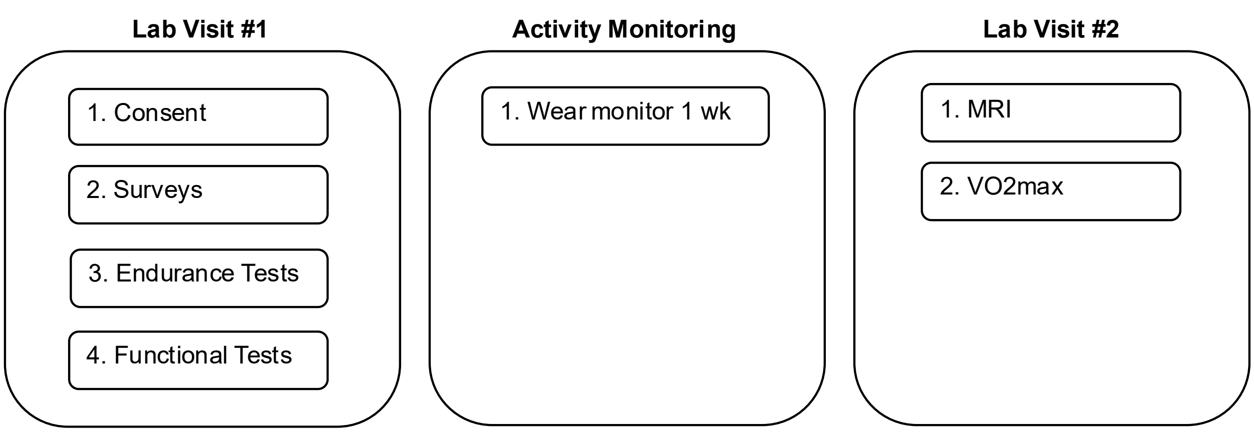

Figure 3.1 – Overview of experimental protocol. Participants attended the lab on two separate days, with at least one week in between to allow for activity monitoring. The lab visits were split into two separate days to minimize the impact of fatigue on the various assessments.

3.3 Participants

In total, 62 participants completed at least Visit #1 of the study. The majority of the participants (42 of 62) were community-dwelling adults aged 50 years or older as changes in muscle mass, strength and power after age 50 are reported to accelerate (Keller and Engelhardt, 2013). Due to their age and typically high levels of PA, masters athletes were also recruited from local Ontario Masters Athletics competitions. To be considered a masters athlete, individuals must "…systematically train for, and compete in, organized forms of competitive sport specifically designed for older adults" (Reaburn and Dascombe, 2008). In sanctioned competitions, masters athlete age groups typically begin at age 35, although this varies by sport with swimming beginning at 25 and golf at 50 years of age (Reaburn and Dascombe, 2008). As a comparison group, a subset of younger adults from general population, and young athletes from York University were recruited. To be considered a (non-masters) athlete, individuals must "…participate in an organized team or individual sport that requires regular competition against others as a central component, places a high premium on excellence and achievement, and requires some form of systematic (and usually intense) training" (Maron et al., 2015). Based on these definitions, all masters athletes were also considered athletes, but not all athletes were considered masters athletes. These questions were posed to participants on intake. A summary of the participants is provided below in **Table 3.1** and a description of the participants who participated in the MRI portion of the study is provided as part of Research Question #5.

Table 3.1 – Description of the participants. Participants shown here participated in Visit #1. A subset of these participants completed the MRI portion (36) and VO_2 portions (36) of Visit #2. Exclusion criteria are explained in section 4.1.

	Overall							
Sample Size	62							
Age (SD)	55.8 (19.2)							
Age Range	22.4 to 85.2							
	Female				**Male**			
Sample Size	23				39			
Age (SD)	54.3 (18.8)				56.7 (19.6)			
Age Range	22.4 to 83.8				23.6 to 85.2			
Height (m)	1.61 (0.05)				1.76 (0.07)			
Weight (kg)	61.2 (10.0)				79.7 (12.2)			
	Athlete			**Non-Athlete**	**Athlete**			**Non-Athlete**
Sample Size	8			15	22			17
Age (SD)	56.2 (18.0)			53.3 (19.7)	56.5 (20.3)			57.0 (19.1)
Height (m)	1.61 (0.06)			1.61 (0.05)	1.74 (0.06)			1.78 (0.06)
Weight (kg)	56.7 (7.6)			63.6 (10.5)	73.8 (11.5)			87.3 (8.5)
	Short	**Long**	**Other**	**Non-Athlete**	**Short**	**Long**	**Other**	**Non-Athlete**
Sample Size	1	5	2*	15	9	6	7*	17
Age (SD)	70.9	63.3 (10.9)	31.2 (8.2)	53.3 (19.7)	64.7 (17.8)	55.4 (24.7)	46.9 (17.6)	57.0 (19.1)
Height (m)	1.60	1.62 (0.05)	1.61 (0.06)	1.61 (0.05)	1.73 (0.05)	1.76 (0.08)	1.74 (0.07)	1.78 (0.06)
Weight (kg)	55.7	52.9 (5.2)	66.8 (4.6)	63.6 (10.5)	74.6 (8.3)	68.4 (11.1)	77.5 (14.8)	87.3 (8.5)
* CrossFit & Field Hockey					* Rugby, CrossFit (2), Middle Distance (2), Field Hockey, Dragonboat			

In addition, An *a priori* power analysis was conducted in G*Power (Faul et al., 2007) with $\alpha = 0.05$ and $\beta = 0.80$. Effects sizes were calculated from regression analyses in existing literature.

In 2004, Runge et al analyzed the effect of age and sex on lower-body power output in a vertical jump. They presented correlation coefficients of 0.81 and 0.86 for females and males, respectively. Based on these data, an a-priori power analysis using F-tests for multiple linear regression with three predictors and an f^2 of 1.90 (converted from r=0.81) determined that a sample size of 11 was required to observe the effect of sex and age on power. Alvero-Cruz et al., 2021 analyzed the effect of age on power output in various athletic disciplines. They presented coefficients of determination of 0.21 for non-athletes, 0.32 for mixed athletes and 0.42 for power athletes. Based on these data, an a-priori power analysis using F-tests for multiple linear regression with three predictors and an f^2 of 0.25 (converted from $R^2 = 0.21$) determined that a sample size of 48 was required to observe the effect of age and athletic discipline on power. In 2017, Hernandez and colleagues evaluated the relationship between PA measures and lower body power in men with COPD. Their analysis showed a correlation of r=0.52 between low intensity PA and lower body power output. No relationship was uncovered between moderate PA and power output. In 2000, Foldvari and colleagues investigated various factors which predicted functional status in community-dwelling elderly women. They found that habitual PA was positively correlated with physical function (r=0.49). Therefore, an a-priori power analysis using F-tests for multiple linear regression with three predictors and an f^2 of 0.32 (converted from r=0.49) determined that a sample size of 35 was required to observe the effect of age and PA on power.

With respect to joint power, the literature is sparser however a study by Garcia and colleagues in 2011 showed a correlation of r=0.44 between questionnaire-based PA level and knee flexor power. Therefore, an a-priori power analysis using F-tests for multiple linear regression with three predictors and an f^2 of 0.24 (converted from r=0.49) determined that a sample size of 50 was required to observe the effect of age and PA on joint power.

Based on the above, for adequate statistical power, a minimum of 50 participants were required to observe the smallest effect of interest, but in the end due to high public interest, 63 participants were recruited (with one drop out).

3.3.1 Inclusion Criteria

Participation screens were typically carried out in advance to avoid participants being excluded from the study after traveling to the campus. Participants first reviewed and signed a consent form which described the procedure presented below, including any minimal risks and benefits associated with their participation. Following the consent form, participants signed a PAR-Q+ questionnaire which helped to ensure that participants were sufficiently healthy to carry out the physical testing and failing a PAR-Q+ was the primary exclusion criteria, in addition to the use of a cane or pregnancy. An MRI screen was also completed to ensure that participants were cleared to undergo MRI at a later date. For Visit #1, no prospective participants were excluded based on the consent form, the PAR-Q+ form, use of a cane or pregnancy. One participant withdrew from the study following Visit #1 for undisclosed personal reasons. For the MRI portion of Visit #2, ten participants were disqualified by the MRI screen most typically due to surgeries involving implantations that could not be proven MRI-safe, 14 participants declined participation for various reasons including claustrophobia and schedule conflicts, and communication was lost with two participants. In total, 36 participants participated in the MRI portion of the study. For the VO_2 evaluation portion of Visit #2, 22 participants declined to participate (no further details were requested), communication was lost with two participants, one was disqualified based on the original PAR-Q+ questionnaire due to recent heart surgery, and one was disqualified based on a medical condition discovered following Visit #1. In total, 36 participants participated in the VO_2 portion of the study. Most of the participants who completed the MRI completed the VO_2 assessments but there were cases where one was completed without the other. The participation screens described above are reproduced in full in **Appendix B**.

3.4 Measures

3.4.1 Questionnaires

Several questionnaires were used to determine the PA and exercise habits of the participants, their overall health and well-being, their PA history and history of injury. All additional detail pertaining to these questionnaires including assessment of their validity is provided here. The questionnaires are reproduced in full in **Appendix C** (PA questionnaires) and **Appendix D** (Custom Questionnaires).

IPAQ-SF

The IPAQ-SF survey assesses the frequency and duration of vigorous PA, moderate PA and walking in addition to sedentary time, taking approximately ten minutes to complete (Lachat et al., 2008). The IPAQ has previously demonstrated acceptable agreement in measuring moderate-to-vigorous PA in older adults, and at least fair validity in measuring sedentary behaviour in the same population (Cleland et al., 2018).

RAPA

The RAPA questionnaire has seven levels of PA to select from, each increasing in duration or intensity from the previous selection. In addition, it asks the user to report (binary yes/no) strength training and/or flexibility training. The RAPA questionnaire has been shown to be a valid measure of PA in older adults (Topolski et al., 2006).

HUNT

HUNT exercise questionnaire is a simple three-question tool developed for longitudinal study of the Norwegian population. It has the benefit of being based on perceived intensity (Kowalski et al., 2012) has been applied to cohorts aged 20 to over 90 years of age (Garnvik et al., 2018), allows for conversion and comparison to WHO guidelines (Bull et al., 2020; Lerfald et al., 2023), and typically takes under two minutes to complete.

RAND-36

RAND 36 survey assesses physical functioning, bodily pain, perceived activity limitations for physical or emotional reasons, emotional well-being, social functioning and general perceptions of ones own health (Hays and Morales, 2001). This survey is commonly used in research on health-related quality of life in older adults (Vetrovsky et al., 2019). It involves 36 questions and takes roughly 7-10 minutes to self-administer (Topolski et al., 2006).

Custom Injury and Physical Activity History Questionnaire

The custom questionnaire contained six sections taking approximately 15 minutes to complete; 1) overuse injuries, 2) acute injuries, 3) specifics related to low-back injuries (Bahr et al., 2004; Fett et al., 2017; Kuorinka and Andersson, 1987), 4) life-long PA history, 5) occupation-related PA and 6) previous surgeries, and current medications. Injury history was of interest due to immediate effects on movement patterns and potential downstream effects on PA habits. This was of particular interest in this cohort as masters athletes are known to have a high prevalence of injury but lower prevalence of chronic disease compared to non-athletes (Patelia et al., 2018). Physical activity history was important to collect as well since recent investigations have begun to clarify the effects of early versus late-starting masters athletics on various physical outcomes such as lean body mass, grip strength, jump power and bone mineral density (Piasecki et al., 2019). This study contributes to that body of knowledge.

3.4.2 Endurance and Control Testing

Several assessments of core endurance and control were conducted to help describe the fitness level of the cohort and also to potentially explain any atypical performance if it occurred. This analysis was not part of a formal research question, however. All endurance tests were capped at a maximum of three-minutes based on the endurance times achieved in an earlier investigation by McGill and colleagues (1999).

Front Plank

Trunk flexor endurance testing was assessed via front plank (**Figure 3.2**) for maximum hold time supported by the elbows and feet (capped at 180 seconds).

Figure 3.2 – Front plank assessment for trunk sagittal flexion endurance.

Side Plank (Left and Right)

Trunk lateral flexion endurance was assessed in the side plank position (**Figure 3.3**) with the top foot in front of the bottom foot per McGill and colleagues (1999), for maximum hold time (capped at 180 seconds).

Figure 3.3 – The side plank assessment for trunk lateral flexion endurance.

Biering Sorenson

Trunk extensor endurance testing was performed using an isometric bodyweight posterior extensor test (Biering Sorenson position; **Figure 3.4**) for maximum hold time, capped at 180 seconds (Brumitt et al., 2013).

Figure 3.4 – The Biering Sorenson assessment for trunk sagittal extension endurance.

AHAbd Test

The active hip abduction task was used to provide an indication of the individual's ability to maintain control of the trunk and pelvis (Nelson-Wong et al., 2009). This test requires a participant to lay on their side with both limbs extended and the pelvis in the frontal plane (**Figure 3.5**, left. From this position, the participant is asked to abduct the hip without bending their leg (**Figure 3.5**, right). The stability in the frontal plane is scored later by an examiner.

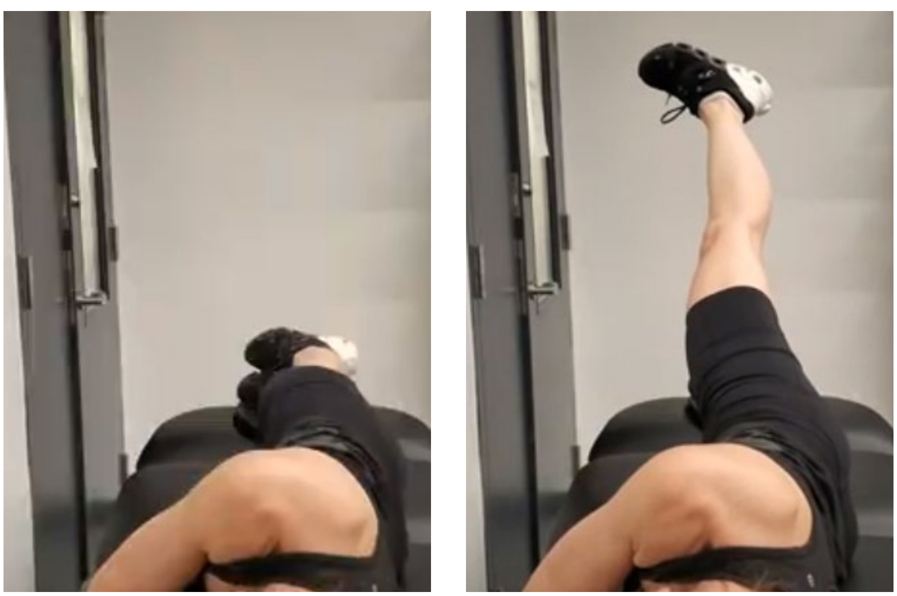

Figure 3.5 – The AHAbd assessment for lumbo-pelvic control.

3.4.3 Functional Testing

Part four of the first visit consisted of the functional testing on the stair apparatus using force plates and motion capture technology described in detail next. Participants performed all jumping, chair rising and stair climbing on the custom-built three-step stair apparatus (**Figure 3.6**).

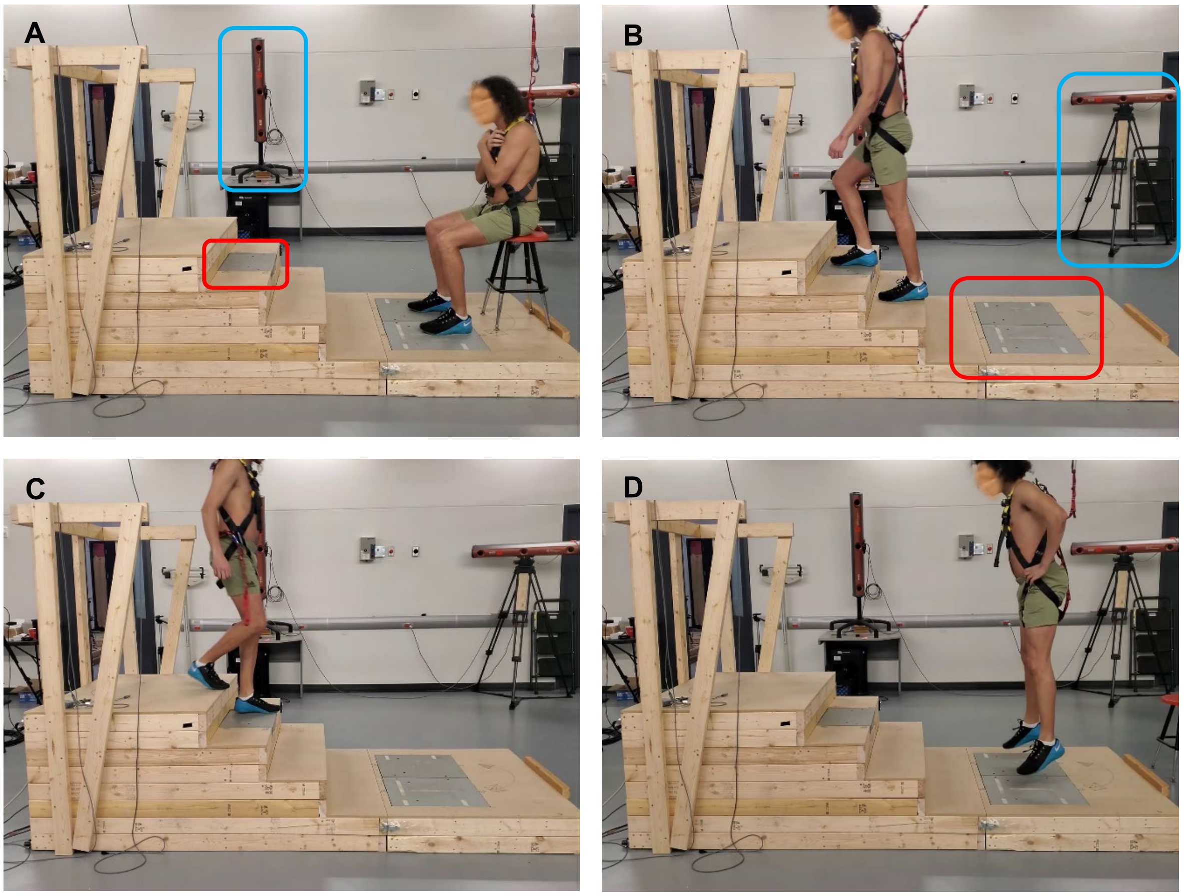

Figure 3.6 – Stair apparatus for functional task assessment. Image A shows the beginning of the chair rise task. From this position, participants stood from the seated position at a casual pace, keeping the arms folded across the chest. Image B and C show the stair ascent and descent, respectively. Participants were instructed to climb the stairs as though they were at home, with no preference to their footing. Image D shows the countermovement jump (CMJ) which, in contrast to the functional tasks, required maximal effort. Three force plates are outlined in red, and two of five motion capture cameras are outlined in blue. Motion capture marker configuration not shown here.

All force plates were removable but fixed in place with the use of set screws contacting the dead end of the force plate. A minimum clearance of 0.5cm was provided between the flush surface of the wooden platform and live end of the force plates. The stair force plate was embedded in the second step to ensure that participants achieved steady state climbing when striking the force plate. The apparatus was outfitted with a railing surrounding the top platform and no handrail was installed as it would disrupt GRF data. Trials with an incomplete foot strike, with the foot partially in contact with wooden surround, were excluded from analysis. The height of the backless chair was set to 56cm and was not adjusted for each participant. The justification was such that most furniture and benches are not adjusted for varying limb segment lengths in everyday life. Thus, a constant chair height may uncover “real life” performance bottlenecks based on uncontrolled factors such as body height. Similarly, the stair height was fixed at 18cm, the depth was 28cm and the width was 130cm.

3.4.4 Kinematics

Motion capture data was acquired at a sample rate of 80Hz, 90% power and 4600Hz strobing frequency using NDI First Principles software (v. 1.2.4) with five NDI 3D Investigator position sensors then modeled and analyzed in Visual3D (HAS-Motion, formerly C-Motion). The marker configuration was modeled after the IOR full body gait model (Cappozzo et al., 1995), but modified for lower body and trunk. This model consists of bilateral feet, shanks, thighs, a pelvis, and a lumbar spine segment using 40 active markers spread over eight rigid bodies in total as shown in **Figure 3.7** below. Four additional active markers were rigidly fixed to the stair apparatus to facilitate digitizing the corners of the force plates, and another four markers were occupied by the digitizing pen itself. In total, 48 active markers were used.

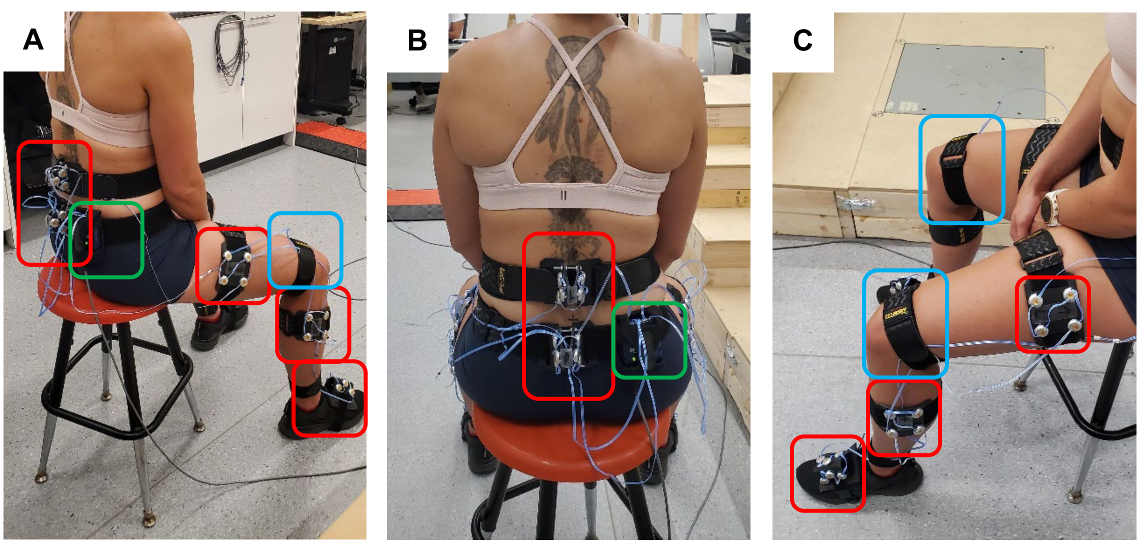

Figure 3.7 – Kinematic marker configuration. Participant shown with eight rigid bodies containing 40 active markers, sampled at 80Hz. Once digitized, this marker configuration generates foot, shank, thigh, pelvis, and lumbar spine segments. All rigid bodies are outlined in red, strober control unit (SCU) is outlined in green and additional straps used to control wires are outlined in blue.

3.4.5 Kinetics

Three AMTI OR6-7-1000 force plates were used for ground reaction force data collection. One force plate was embedded in the stair apparatus, and two were embedded in the lower platform. All force plates were sampled at 1200Hz and synchronized with the 80Hz motion capture data via NDI First Principles. The stair force plate connected to a 6-channel amplifier (MSA-6 MiniAmp), which connected to a 16-channel BNC Breakout Box (Advanced Systems), which connected to an ODAU unit (NDI, ODAU II), which finally connected to the System Control Unit (NDI, SCU). The floor force plates had a similar series of connections, although these two plates shared a single 16-channel BNC breakout box (using 12 of 16 channels). Note that the first 38 participants' data were collected using a single force plate embedded in the lower platform. The last 25 participants' data were collected with the correct configuration of two plates embedded in the lower platform. The implications of this discrepancy are analyzed and discussed in **Appendix G**.

3.4.6 Cardiorespiratory Testing

$VO_{2,max}$ was measured during a graded exercise test using a PNOE portable metabolic analyzer (PNOE, California, USA) as an incentive for the athlete population, and to further characterize the fitness of the participants. This system has previously been validated against the COSMED metabolic cart, showing very strong correlations for VO_2 (r=0.98), VCO_2 (r=0.98), VE (r=0.98) and RQ (r=0.91(Tsekouras et al., 2019). The average difference in gas flow rates measured by the PNOE system versus the COSMED system was approximately 34.0 ± 118 ml/min for VO_2 and 36.4 ± 110 ml/min for CO_2 (Tsekouras et al., 2019). A StarTrac treadmill was used which permitted a maximum speed of 19.3 km/h (12 mi/h) and 15% grade. If requested, participants were allowed to use a full body fall arrest harness during the assessment (one such occurrence). Details on the ramp protocol are provided in the procedure section.

3.5 Procedure

3.5.1 Questionnaires

Participants began their visit by filling out paper-format questionnaires including the IPAQ-SF, RAPA, HUNT, RAND-36, and the custom questionnaire. In total, all questionnaires took approximately 30-40 minutes to complete. All questionnaires are reproduced in full in **Appendix C** (PA questionnaires) and **Appendix D** (custom questionnaires).

3.5.2 Endurance and Control Testing

Following questionnaires, participants underwent trunk endurance and pelvic-control testing. All endurance testing was performed within a three-minute time-cap based on normative data from McGill and colleagues (1999), with no verbal encouragement provided. Following each effort, all participants were given a rest period equal to their work period. Participants were invited to warm up at a self-selected pace on a treadmill (Star Trac ETR) or cycle ergometer (Star Trac SUBX). Following a brief warm-up, they began the assessments listed below.

Front Plank

Participants performed the front plank testing with the elbows beneath the shoulders and a straight line formed in the sagittal plane through the shoulders, hips, knee and ankles. Any deviation from this position received a warning, with any subsequent deviation ending the test. Thirty eight of 62 participants reached the three-minute time-cap on this assessment, and 29 of this 38 were over the age of 50.

Side Plank

The side planks were conducted similarly with the elbow beneath the shoulder and the trunk and lower limbs being arranged linearly. The foot of the top leg was permitted to be on the ground in front of the bottom foot (as opposed to stacked on the bottom foot) to improve stability. Similar to the front plank, any deviation from this position received a warning, with any subsequent deviation ending the test. Six of 62 participants reached the three-minute time-cap for the left-side assessment, five of whom were over the age of 50. Five of 62 participants reached the three-minute time-cap on the right-side assessments, four of whom were over the age of 50.

Biering Sorenson

The Biering Sorenson test was conducted in the prone position on a physiotherapy plinth with the anterior superior iliac spines (ASIS) at the edge of the plinth leaving the upper extremities cantilevered and unsupported. Prior to the test beginning, participants were permitted to rest on a stool which was placed under their torso for support. Once they were prepared, the stool was removed, and the participant raised their torso through back extension until parallel with the ground (Brumitt et al., 2013), with arms folded across the chest. Any deviation from this position received a warning, with any subsequent deviation ending the test. Five of 62 participants reached the three-minute time-cap on this assessment. Fourteen of 62 participants reached the time-cap on this assessment, 12 of whom were over the age of 50.

AHAbd

Following the endurance testing, the participant was invited to perform the Active Hip Abduction Task (AHAbd). Prior to making their attempt, a research team member demonstrated the correct execution of the assessment. Then the participant assumed a side-laying position and was provided the following verbiage:

> “Please keep your knee straight and raise your top thigh and leg towards the ceiling, keeping them in line with your body, and try not to let your pelvis tip forwards or backwards.” (Nelson-Wong et al., 2009)

The participant was not allowed to practice the movement. The assessment was recorded to video for later assessment by multiple raters.

3.5.3 Functional Testing

Once configured for motion capture, participants completed two repetitions of the chair rise task. Prior to the task, the following instructions were provided to each participant:

1. “Keep your arms folded across your shoulders”

2. “Stand up casually as though you’re at home, standing from the kitchen table”
3. “Once standing, please stay standing, until I tell you to sit”

Following two chair rise tasks, participants completed two stair climb and stair descents. The following instructions were provided:

1. “Keep your arms across your shoulders”
2. “Climb the stairs naturally”
3. “Use whichever foot you would like, but please ensure that you strike the force plate”

Following the chair rise and stair tasks, participants were instructed briefly on the CMJ assessment which involved a countermovement to the participants’ desired depth, followed by a maximal effort reversal of direction (a jump). Instruction consisted of the following:

1. “Keep your hands on your hips at all times”
2. “Try to jump as high as possible”
3. “You do not need to land on the force plates”

Following instruction, participants were invited to complete a minimum of three warmup CMJ attempts at 25%, 50% and 75% perceived effort. Some participants, typically those who had never jumped for the express purpose of exercise, would stomp the ground upon landing. In this circumstance, one additional cue was presented to participants: “Try to land gently”. Nonetheless, existing research has shown that in an athletic population, vertical jumps do not likely need a familiarization session (Moir et al., 2005). Participants were then invited to complete three maximal attempts with 20 seconds of rest between each effort. During the CMJ task, hands were held on hips for several reasons; 1) to limit the likelihood of disconnecting motion capture wiring during arm swing, 2) to limit upper body contributions to power (not included in modeling), and 3) to limit differences in technique (Korhonen et al., 2009). The jump was performed on force plates surrounded by additional platforming at a consistent height to ensure safe footing. At least one additional lab member was standing by to assist in the event of a fall.

The final component of functional testing involved a 15-minute functional task circuit. One round of this circuit involved 3x chair rise, stair ascent / descent, 3x chair rise, stair ascent / descent and 1x CMJ (**Figure 3.8**). Power output during the CMJ was used assess lower body fatigue at the end of each round. Participants were asked to complete as many rounds as possible in 15 minutes. The functional tasks were done at a casual pace, and the countermovement jump was performed with maximal effort. The circuit was designed to maximize the quantity of functional task repetitions and to minimize the quantity of CMJ repetitions. The CMJ was not meant to induce fatigue, but rather it was meant to measure fatigue that arose from functional task performance. For this reason, two rounds of functional tasks were performed before one CMJ. During pilot studies, different versions of the circuit with a higher ratio of functional tasks were performed, but there was concern that participants with low ability would take more time to complete the functional tasks, thereby performing less CMJs, which would have compromised the fatigue analysis (it requires several CMJ).

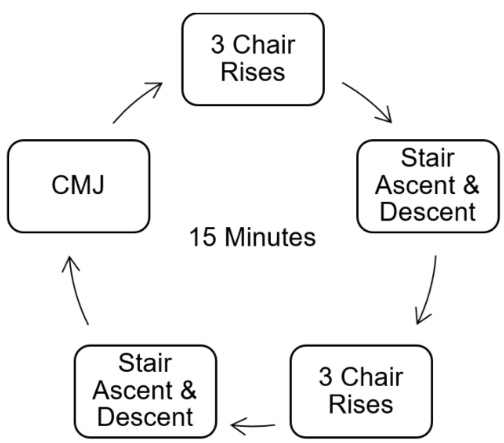

Figure 3.8 – Overview of the functional task circuit. Participants began the circuit with three chair rises and continued through the circuit for 15 minutes.

The instructions for the functional task circuit provided to the participants were as follows:

1. “Perform the chair rise and stair task casually, as though you are at home.”
2. “The CMJ, on the other hand, must be performed with maximal effort.”
3. “Rest as needed”

3.5.4 Cardiorespiratory Testing

The treadmill-based $VO_{2,max}$ protocol began with PNOE device calibration to outdoor air. Following this calibration, the mask and backpack were fitted to the participant. Once the mask was secured, a fit test was conducted by obstructing the outlet and exhaling forcefully. The mask was deemed to fit well if little to no air leaked from the mask. All participants were reminded that they could stop the test at any time, and for any reason. For participants with no running experience (treadmill or otherwise), or those who were uncomfortable running, a graded walking test was conducted (5 older adults, of 36 total VO_{2max} assessments). In the first four minutes, participants warmed up between 3.2 to 4.8km/h (2 and 3 mi/h) and 1% grade. For some, their working speed was achieved at the end of this four-minute period and if the participant agreed to continue, the step test began, and the grade was increased to 3%. At each two-minute interval, if the participant agreed to continue, the grade was increased by 2% until one of the following endpoints was reached:

1) $VO_{2,max}$ reached: VO_2 did not change between successive two-minute bouts, *and* the participant reached volitional exhaustion.
2) $VO_{2,peak}$ reached (incomplete test): Participant reached volitional exhaustion.

In masters athletes, younger subjects, and/or those comfortable with running, the protocol was similar, differing only in the treadmill speed. Following the four-minute warmup period, speed was increased by 1.6 to 2.4 km/h (1 or 1.5 mi/h) every two minutes until a comfortable pace was achieved. At this point, the step test began, where grade was increased by 2% every two minutes until the same endpoints were met. The fastest comfortable pace was 7mph (11.3kph) and the maximal treadmill gradient was 11% (not achieved by the same participant). The VO_2 test was the final component of the protocol which is summarized below in **Figure 3.9**.

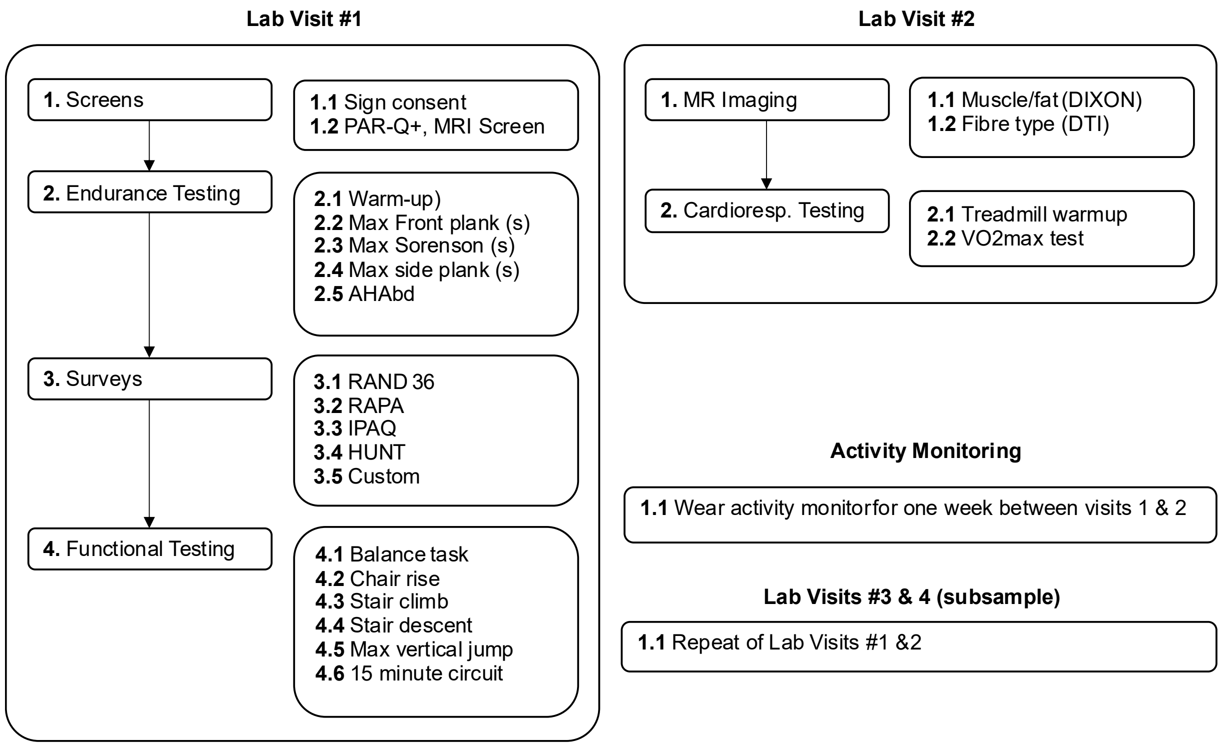

Figure 3.9 – Overview of full protocol. The lab protocol involves physical activity and quality of life surveys, MR imaging for cross-sectional area, fat content, and muscle fiber type approximation. Maximum lower body power, and endurance testing of the torso are assessed next. Functional testing consists of a 15-minute circuit containing a vertical jump to assess power decline (a direct measure of fatigue), a stair climb, stair descent and 3x chair rise task. To finish, participants complete a $VO2_{max}$ test.

3.6 Data Processing

3.6.1 Introduction

The data from questionnaires, 3D kinematics from motion capture cameras, 3D kinetics from force plates and MRI DICOM images required additional processing prior to analyses. The specific programs and processing steps are outlined in this section.

3.6.2 Questionnaires

All questionnaires were completed in paper format and tabulated in Microsoft Excel (v2403). RAND 36 was completed in paper format and graded using Microsoft Excel in accordance with the guidelines published by the RAND corporation. This process involved transferring numeric scores from the paper copy to Excel, then recoding questions into percentage scores, then averaging the percentage scores to generate the scores for the eight health concepts per RAND. The RAPA survey was graded using Microsoft Excel in accordance with the guidelines published by the University of Washington (Topolski et al., 2006). This survey was used to classify older adults as underactive (score of 1-5) or active (score of 6 or 7). The HUNT survey was conducted and analyzed in accordance with Lerfald and colleagues' work (Lerfald et al., 2023). This PA survey was extracted from Norwegian population study, which has been applied to individuals from across a wide age range, from 20 to over 90 years (Garnvik et al., 2018). An additional benefit of the HUNT survey over the RAPA and the IPAQ survey is that the results may be translated into the terms of the WHO guidelines which allows for interpretation on activity level (Lerfald et al., 2023). Both the IPAQ and the RAPA survey have their own activity level classifications, but they do not agree with the current WHO guidelines, specifically with the requirement for moderate or vigorous PA and the 10-minute minimum duration (no longer required). **Table 3.2** below provides more detail surrounding this analysis. The HUNT survey contained the following three questions:

> **Question 1**: 'How often do you exercise?' with the response alternatives 'never' (0), 'less than once a week' (0), 'once a week' (1), 'two to three times per week' (2.5), and 'almost every day' (5).
>
> **Question 2**: 'If you do such exercise as frequently as once or more times a week, how hard do you push yourself ?' with the response alternatives 'I take it easy, I don't get out of breath or break a sweat (1), 'I push myself so hard that I lose my breath and break into a sweat' (2) and 'I practically exhaust myself' (3).
>
> **Question 3**: 'How long does each session last?' with the response alternatives 'less than 15 min' (0.1), '15–29 min' (0.38), '30 min to 1 h' (0.75), and 'more than 1 h' (1.0).

The final PA score was then calculated by multiplying the score of each individual question (the number in brackets) for all three questions, in Microsoft Excel. This continuous score, ranging from 0 to 15, was used for regression analysis in RQ #1 and RQ #2. For categorical analysis (RQ #3), the method described by Lerfald and colleagues (Lerfald et al., 2023) was used to convert continuous scores into weekly minutes of activity done at either a low, moderate or high-intensity. Question 1 (above) determined weekly exercise *frequency*, Question 2 determined weekly exercise *intensity,* and Question 3 determined weekly exercise *duration*. These scores were compared to current WHO guidelines and then segregated into active or underactive categories. An example calculation is provided below for a fictitious participant who provides the following responses:

> **Question 1**: Two to three times per week' (2.5 days, score of 2.5)
>
> **Question 2**: I take it easy; I don't get out of breath or break a sweat (low intensity, score of 1)
>
> **Question 3**: 15-29 minutes (30-minute duration, score of 0.38)

The calculation is carried out as follows:

> 2.5 days per week (frequency) * 1 (intensity) * 30 minutes per day (duration) = 75 minutes of low-intensity PA per week.

Based on this calculation, this participant would receive a score of 0.95 as a continuous variable and an 'underactive' classification for the categorical analysis based on not meeting the minimum of 150 moderate minutes per week.

IPAQ-SF survey scores will be transferred to an automatic scoring spreadsheet (Hoi Lun Cheng, 2016), which tabulates scores for all participants individually in accordance with published IPAQ guidelines for data processing and analysis (IPAQ Group, 2005). Most of the calculation done by the spreadsheet involves trivial multiplication and addition (i.e., minutes of activity per day multiplied by days of activity per week). However, the spreadsheet does also truncate all activities that exceed 180 minutes, in accordance with the published guidelines. This is intended to prevent misclassification into the high activity level category.

In the IPAQ survey, activity level scores can be presented either as categorical variables (1-low, 2-moderate, 3-high) or as a continuous variable of activity-related metabolic equivalents (METs) per week. There are two ways to be classified as low-activity, three ways for moderate-activity and two for high-activity. **Table 3.2** below summarizes all the ways in which a participant can be classified into a given activity level.

Table 3.2 – Examples of possible pathways leading to various activity level classifications. Asterisks indicate pathways which may lead to overclassification of moderate and high activity levels via walking activities.

	Walking (recreation, sport, leisure)		**Moderate** (carrying loads, riding bicycle)		**Vigorous** (heavy lifting, aerobics class)		**Notes**
Classification	**Days**	**Duration**	**Days**	**Duration**	**Days**	**Duration**	**-**
1 - Low	-	-	-	-	-	-	No activity reported
1 - Low	-	-	-	-	2	20	Failure to meet moderate criteria
2 - Moderate	-	-	-	-	3	20	-
2 - Moderate	2	30	3	30	-	-	30 min per day, 5 days any activity combo
*2 - Moderate	5	36	-	-	-	-	600 MET-mins per week any activity combo, 5 days
3 - High	-	-	-	-	3	63	1500 MET-mins per week vigorous activity
*3 - High	7	130	-	-	-	-	3000 MET-mins per week any activity combo, 7 days

MET minutes are calculated in the spreadsheet in accordance with the IPAQ guidelines, which itself utilizes MET values from the 2011 Adult Compendium of Physical Activities (Shephard, 2012). One MET is approximately 3.5mL of oxygen consumption per kg bodyweight per minute and equates roughly to the oxygen consumption of an adult in a resting state. The IPAQ thus considers walking to require 3.3 METs on average (or 3.3-fold energy consumption compared to resting), moderate activities to require 4 METs on average and vigorous activities to require 8 METs on average (IPAQ Group, 2005).

Weekly moderate and vigorous PA recommendations can be used to synchronize activity classification from the various surveys. As of 2020, the World Health Organization (WHO) recommends 150-300 minutes of moderate PA per week or 75-150 minutes of vigorous PA per week (Bull et al., 2020; Jakicic et al., 2019). These data can be extrapolated from the activity monitors but also from the IPAQ, RAPA and HUNT surveys (Lerfald et al., 2023).

3.6.3 Kinematics

To facilitate model-building, 29 digitized points were identified on every participant (**Figure 3.10**) and an additional four points were located on each force plate (three plates, 12 points total) to locate the force plates precisely in the model-space (not shown). Data were exported from First Principles as

C3D files and imported into Visual3D for modeling, processing, and analysis. When presented separately, kinematic data were processed by gap-filling to 10% of the sampling frequency, i.e., eight points, then a fourth-order bidirectional low-pass Butterworth filter with a cut-off frequency of 10Hz was applied.

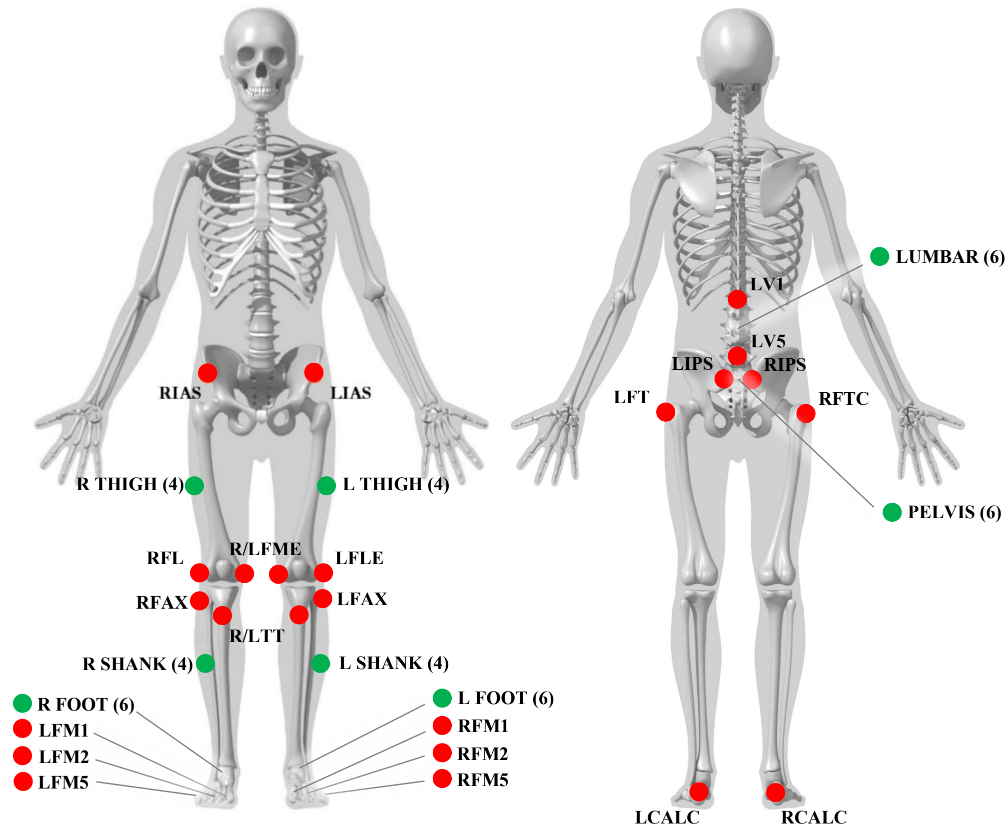

Figure 3.10 – Lower body and lumbar spine marker placement. Red dots • indicate digitized points, while green dots • indicate rigid bodies with the number of markers in parentheses. Image modified from HAS Motion.

Joint angles were interpreted in accordance with Visual3D convention and the right-hand rule, which dictates that all segment coordinate systems are configured with X+ posting laterally (right side from participants perspective), Y+ anterior and Z+ superior. Accordingly, ankle dorsiflexion is positive, knee extension is positive, hip flexion is positive, and trunk extension is positive. **Figure 3.11** below shows a brief example of typical joint angles observed during the CMJ. Events were used in Visual3D to establish the critical phases of the various movements, which allowed for consistent processing within and between participants. The chair rise, stair ascent and stair descent events were set allow for processing of a full gait cycle, i.e., from toe off to toe off. These events are detailed in the following sections.

Figure 3.11 – Illustrative example of the joint coordinate system and corresponding joint angle changes during the CMJ. Yellow lines indicate the reference segment.

CMJ Event Processing

Kinematic triggers were used to set events for the CMJ task (**Figure 3.12**). The vertical ground reaction forces were summed to create a single resultant ground reaction force solely for the purpose of event processing. Every CMJ began with participants standing still on the force platforms for at least 0.5 seconds. This period was sampled for the average height of the CoM of the pelvis segment. The CMJ start event was set when CoM lowered by 0.5% of body height (typically approximately 1cm) from the initial height of the CoM. This technique was used instead of using the force signal as the ground reaction force signal differed not only in magnitude, but also in shape across participants [4], and so a consistent force-derived trigger was not suitable. The CMJ end event was set to when the CoM returned to this same height upon landing. The vertical displacement of the pelvis was used to calculate CMJ height (Dias et al., 2011; Knihs et al., 2021; Leard et al., 2007).

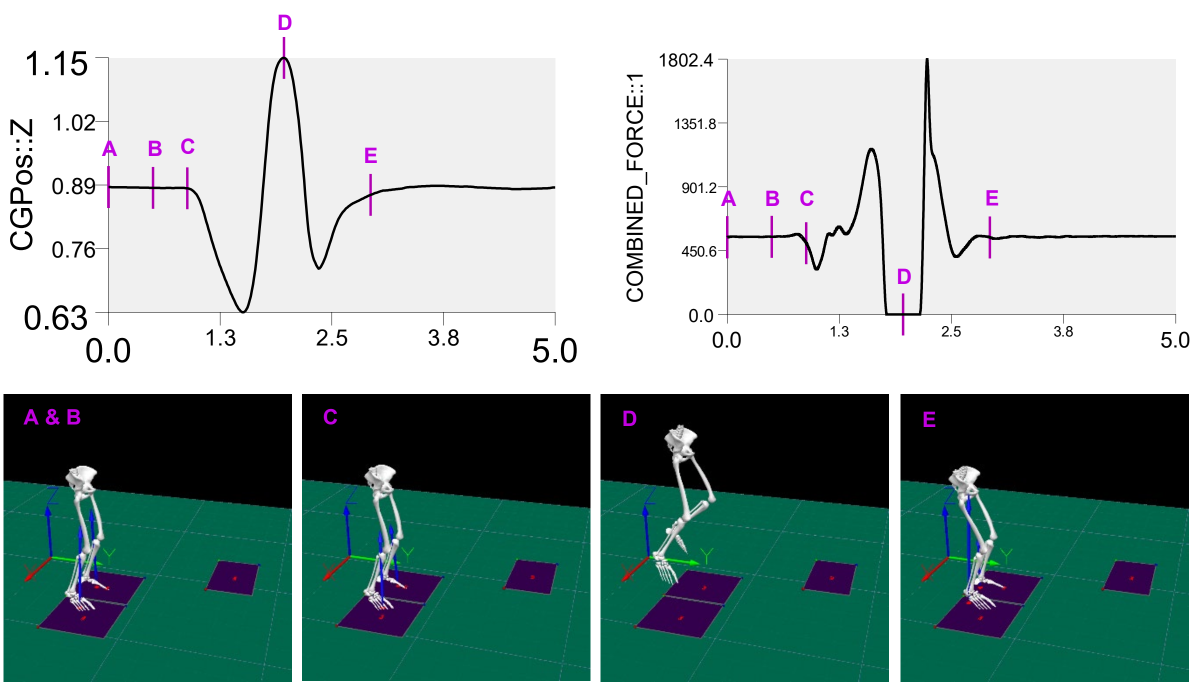

Figure 3.12 – The kinematic measures which were used to trigger events in the countermovement jump (CMJ). The start CMJ event (Event C) threshold was triggered when the centre of mass (CoM) of the pelvis dropped more than 0.5% of the participant's body height (typically approximately 1cm). The end CMJ event (Event E) was triggered when the CoM crossed back through the same height that initially triggered the start event. The peak height of the pelvis CoM is shown in Event D and is used for Event E to determine that the jump has been completed.

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[4] Older adults typically showed a muted countermovement which resulted in a non-distinct unweighting period (force dropping below body weight prior to reversal of direction).

Chair Rise Event Processing

Kinetic and kinematic triggers were used to create events for the chair rise task (**Figure 3.13**). The vertical ground reaction forces were summed to create a single resultant ground reaction force solely for the purpose of event processing. Every chair rise task began with the participant's feet on the force plates while they were seated in the chair. The chair rise cycle began when the resultant force signal surpassed 25% bodyweight (Event A). The end of the task was determined when the vertical velocity of the centre of mass (low pass filtered at 2Hz) crossed zero (Event C). The chair rise peak GRF was determined as the local maximum force between events A and C.

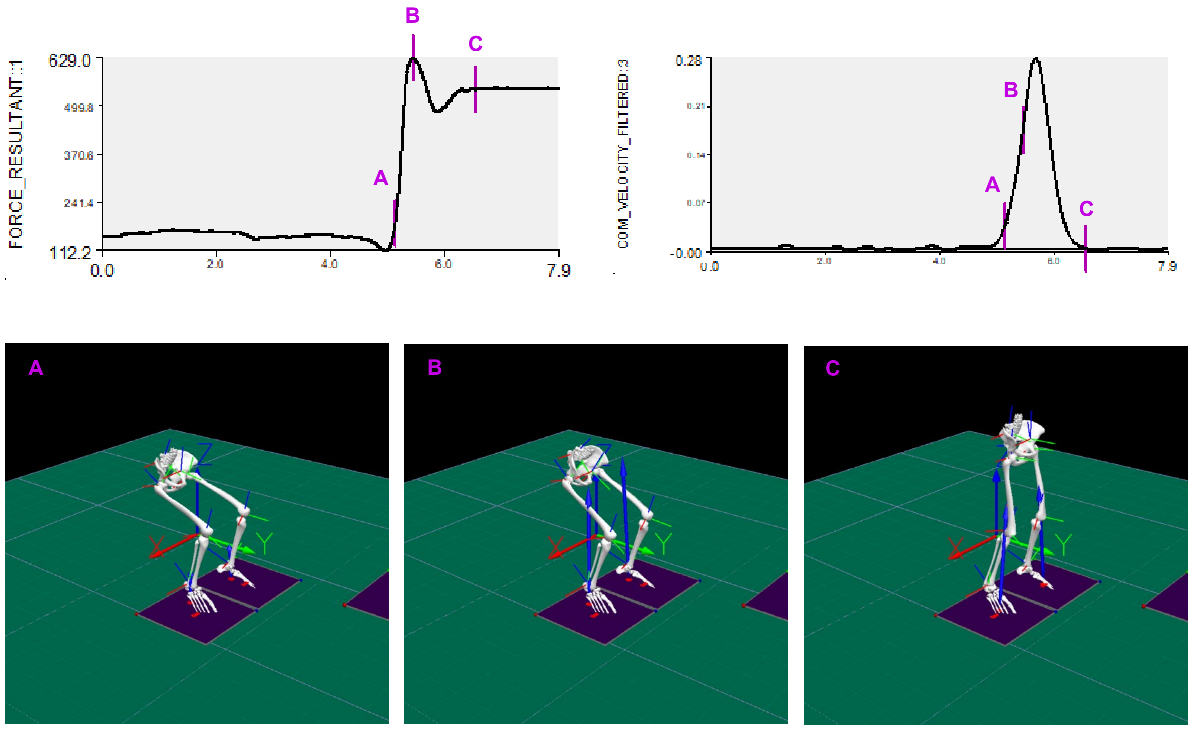

Figure 3.13 – The kinetic (top left) and kinematic measures (top right) which were used to trigger events in the chair rise task. The start chair rise event (Event A) threshold was triggered when the filtered (10Hz) vertical ground reaction force signal surpassed 25% of bodyweight. The end chair rise event (Event C) was triggered when the filtered (2Hz) vertical velocity of the centre of mass achieved stationarity upon standing (first zero-crossing after standing). The peak chair rise GRF (Event B) was determined as the local maximum GRF between events A and C. The biomechanical position represented by each event is shown in the bottom three images.

Stair Ascent Event Processing

Events within the stair ascent gait cycle were also triggered by kinetic and kinematic thresholds (**Figure 3.14**). The stair ascent event (Event A) was triggered when the distal end of the striking foot was greater than 0.03m in the z-direction. Across participants, the distal end of the foot was typically near 0.02m when standing on the platform. Therefore, 0.03m was used as it allowed for some measurement error, and individual variation in shoe stack height. The initial contact event (Event B) by the striking foot was triggered when the force signal was greater than zero. The stair ascent ended when the stair force signal was zero (Event C). Note that a kinematic trigger (Event A) was used to begin the stair ascent, as opposed to beginning at initial contact (Event A) to capture a full toe-off to toe-off gait cycle.

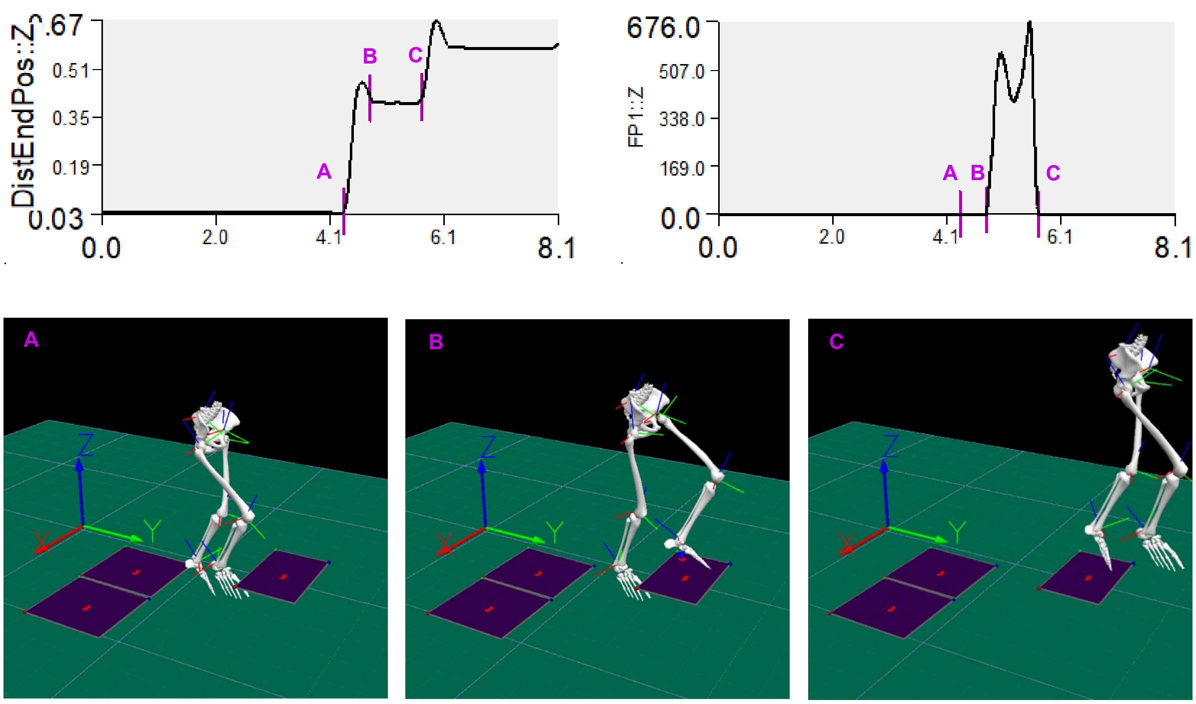

Figure 3.14 – The kinematic (top left) and kinetic measures (top right) which were used to trigger events in the stair climb task. The start stair ascent event (Event A) was triggered kinematically when the distal end of the foot surpassed 0.03m in the z-direction. The initial contact of the striking foot (Event B) was triggered kinetically when the filtered (10Hz) ground reaction force signal surpassed zero Newtons. The end stair ascent event (Event C) was trigger kinetically when the GRF returned to zero Newtons.

Stair Descent Event Processing

Events within the stair descent gait cycle were also triggered by kinetic and kinematic thresholds (**Figure 3.15**). The stair descent start event (A) was triggered kinetically when the ground reaction force was greater than zero Newtons. The stair descent toe-off event (C) was triggered kinetically when the GRF fell to zero Newtons. The stair descent peak GRF event (B) was determined as the maximum GRF between Events A and C. Finally, the end stair descent event (D) was triggered when the distal end of the foot fell below 0.03m in the z-direction. As mentioned above, a position of 0.02m corresponded to the average position of the distal end of the foot in the z-direction when standing on the lower platform surface. This method captures a full gait cycle from initial contact on the force plate to initial contact of the striking foot on the platform.

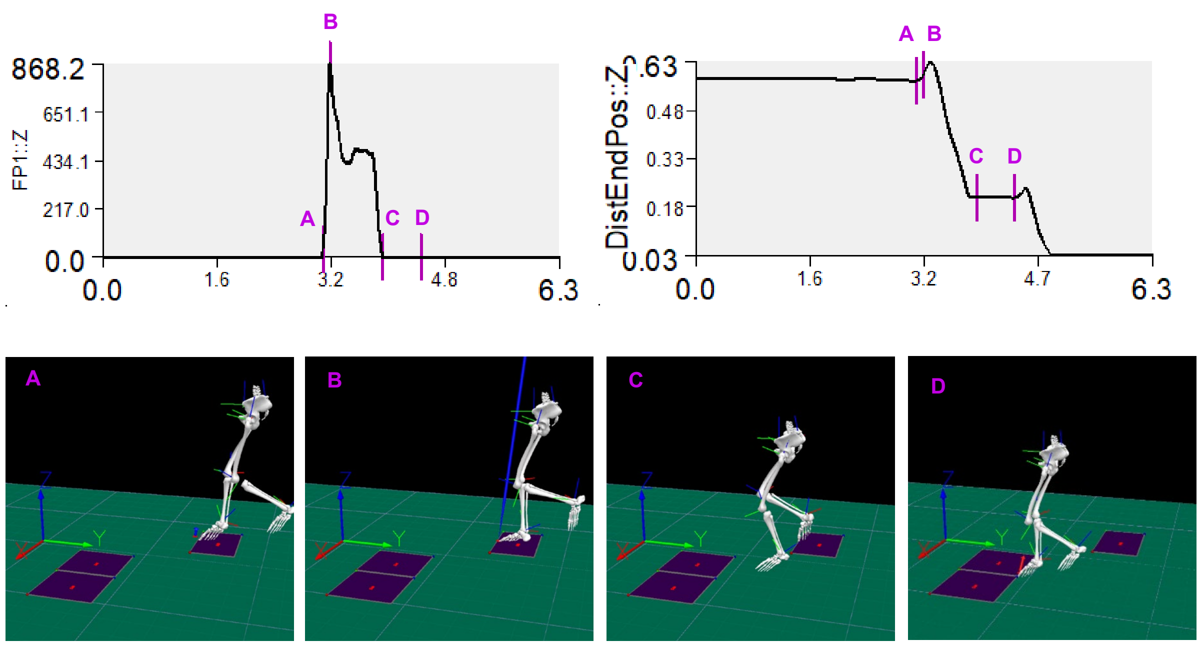

Figure 3.15 – The kinetic (top left) and kinematic measures (top right) which were used to trigger events in the stair climb task. The start stair ascent event was triggered kinetically when the filtered (10Hz) ground reaction force signal rose above zero (Event A). The toe off event (C) was triggered when the stair GRF fell below zero Newtons. The maximum GRF event (B) was determined as the maximum GRF between events A and C. The end stair ascent event was trigger kinematically when the distal end of the foot approached to within 0.03m of the floor (Event D).

Circuit Event Processing

The circuit consisted of three chair rises, a stair ascent and stair descent, repeated twice, followed by one CMJ to finish the round. The chair-rise start event was triggered when the combined force signal exceeded 25% bodyweight. This value protected against error due to varying foot weight on the force plates across participants. Events in the chair rise, stair ascent and descent were calculated as discussed above. The only differences in the circuit were related to choosing which chair rises and stair trials to analyze. For the chair rise task, the first and fourth chair rises (of six total in each round) were analyzed. These repetitions were the most consistent from a kinetic standpoint since they were begun from a resting position (**Figure 3.16**).

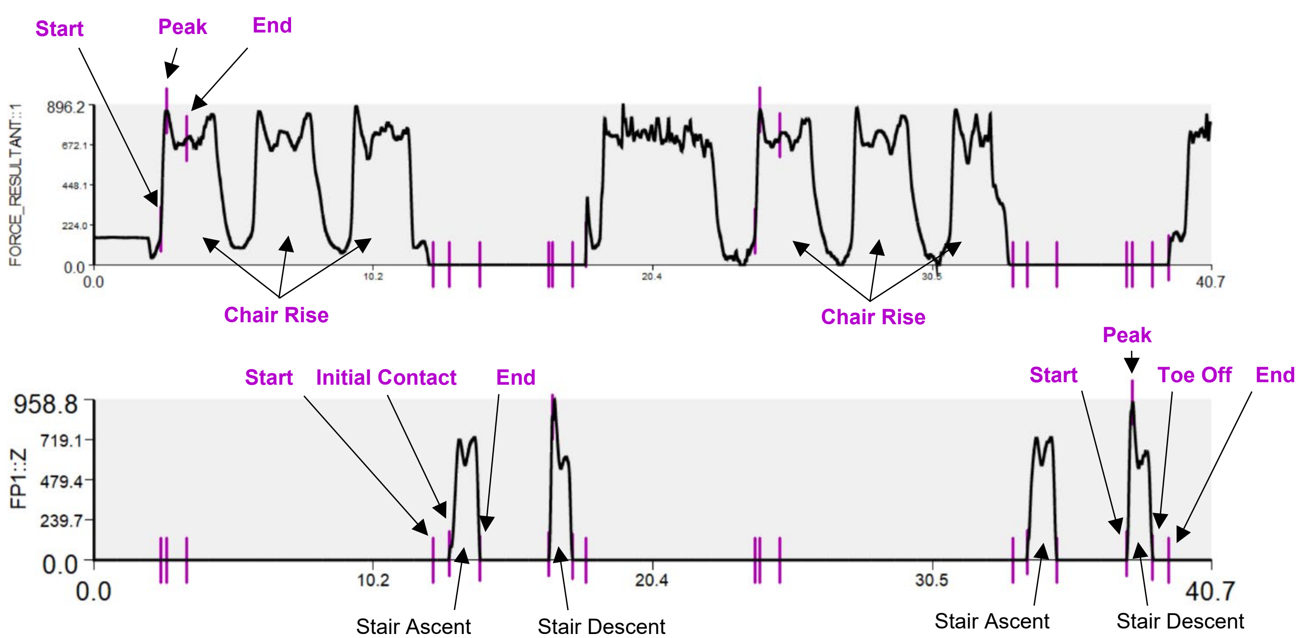

Figure 3.16 - Depiction of event analysis in the functional task circuit. The Pre and post conditions are both comprised of two chair rises, two stair ascents and two stair descents as shown above. The first and fourth chair rises of each circuit were analyzed as they showed the most consistent kinetic data.

Analysis for the stair ascent and descent was conducted on similar foot strikes, i.e. a left-footed ascent at pre to a left-footed ascent at post. The implication of this is that some trials were excluded from analysis even if they were truly earlier in the circuit than the round selected to represent the pre condition, and vice verse for the post condition. However, on average participants fell into a rhythm quickly and across 60 participants analyzed for this particular outcome, the two selected circuits bookended 89% of the total circuits completed. To clarify this figure, 89% implies that with 19 rounds being completed (for example) by the participant, stair ascent and stair descent analysis would be conducted at round three and 19 (or two and 18, etc.). Practically speaking, excluding rounds near the beginning was more common as participants typically found a rhythm early on and continued that foot strike pattern throughout.

Using the previously mentioned events, movement strategies were investigated in both the CMJ and the functional task circuit near the beginning of the 15-minute circuit (the pre condition) and near the end of the circuit (the post condition). In the CMJ, joint angles were computed at the instant of peak force and peak (whole body) power. For the chair rise and stair descent, joint angles were computed at the peak ground reaction force (GRF). For the stair ascent task, the initial contact (IC) event was used to compute joint angles for the sake of consistency. The GRF signal during the stair ascent was bimodal and varied within and between participants in terms of which peak of the two showed the maximum GRF. RMSD was used to determine which joints angles changed the most, regardless of directionality (flexion versus extension).

3.6.4 Kinetics

For inverse dynamics analysis, the lower body and trunk will be modeled as a series of linked-segment rigid bodies. Mass, centre of mass location and moments of inertia were as default in Visual3D, which draws data from several well-accepted sources (Dempster, 1955; Hanavan, 1964; Plagenhoef et al., 1983). The model assumes joints act as pure hinges (i.e., no translation), that joints offer no friction, and segments experience no air-resistance. Newtonian equations of motion were used to calculate joint reaction forces and moments. The participant’s height and weight (measured same day) were entered into the Visual3D model and no other model parameters were modified in Visual3D. As mentioned previously, four corners of all force platforms were digitized in kinematic space to allow translation of kinetic data into the kinematic reference frame. Force plate calibration matrices were obtained and entered in SI units into Visual3D.

Kinematic data was filtered with a fourth order, bi-directional, low-pass Butterworth filter with a cut-off frequency of 10Hz (Pupo et al., 2013; Vlietstra, 2014). Kinematic and force plate data were both filtered with the same filter technique and cut-off frequency prior to feeding the data into the Visual3D model (Bisseling and Hof, 2006).

Lower-body specific power during the CMJ was evaluated using force and kinematic data. Peak lower-body power was calculated as the greatest product of the vertical velocity of the pelvis CoM and net GRF before the moment of takeoff (**Equation 3.1**). The vertical velocity of the pelvis CoM is a standard output form Visual3D and was used as an analogue to whole body CoM since only the lower body was modeled, and therefore only a lower body CoM could be solved for within Visual3D. Net vertical force was calculated as total vertical force (both force plates) minus bodyweight in newtons (**Equation 3.2**).

Equation 3.1 $$\text{Lower Body Power} = F_{net_Z} * V_{pelvis_Z}$$

Equation 3.2 $$Where, F_{net_Z} = \left(F_{2_Z} + F_{3_Z}\right) - Bodyweight\ (N)$$

3.7 Data Analyses

Statistical analysis was conducted with linear regression, ANOVA and confidence intervals. For each regression analysis, linearity was first confirmed via scatter plot of the DV versus each IV. Normality was verified via visual observation of the standardized residual histogram and p-p plot. Noncollinearity was confirmed with a variance inflation factor (VIF) below 10 and statistical independence confirmed via Durbin Watson statistic between 1.5 and 2.5. Effect size was reported via coefficient of determination, R^2. Outliers outside of three SDs from the mean were removed from the analysis. The one occurrence is discussed in detail in the Methods section of Research Question #1. For ANOVA analyses, normality was assessed by visual inspection of p-p plots, and homogeneity of variance was tested via Levene's Test of Equal Variance. Significant interactions were reported when found and main effects were reported only in the absence of significant interactions. Effect size was reported via partial eta squared and post-hoc testing was conducted with Bonferroni correction when a significant effect was found. Confidence intervals were based on normal distributions with exception to Hypothesis 5.2 which used a t-distribution due to smaller sample size. All statistical testing employed a significance level of $\alpha = 0.05$. A brief description of the analyses for each research question is provided below, and a detailed description is provided in each chapter.

Research Question #1

Hypothesis 1.1 determined if age and activity level made significant contributions to three multiple linear regression models of lower body specific power during the CMJ. One instance generated a model with all participants with age, activity level and sex as IVs, while two additional regression analyses generated models for each sex, with age and activity level as IVs. Hypothesis 1.2 determined if age and activity level made significant contributions to 12 multiple linear regression models of joint power during all functional tasks. A model was generated for each joint (ankle, knee & hip) and each task (CMJ, chair, ascent, descent).

Research Question #2

Hypothesis 2.1 was tested determined if age and athletic discipline made significant contributions to a multiple linear regression models of lower-body specific power during the CMJ. One regression analysis generated a model for all participants with age, sex and athletic discipline as IVs, and four additional regression analyses generated models for each athletic discipline (short, long, other and non-athlete) with age and sex as IVs. Hypothesis 2.2 determined if age and athletic discipline made significant contributions to a predictive model of joint power during all functional tasks and was tested via 12 instances of multiple linear regression. A model was generated for each joint (ankle, knee & hip) and each task (CMJ, chair, ascent, descent).

Research Question #3

Hypothesis 3.1 assessed the effect of age sex and activity level on the change in lower body specific power via a single three-way ANOVA. A 95% CI was used to determine if the change in power was significantly different from zero. Additional inquiry into kinematics used boxplots, 95% Cis and descriptive statistics such as coefficient of variation (SD/μ) which can be used to describe the spread of data. In biomechanics, a CV of 0.05 describes a narrow distribution, whereas a CV of 0.5 or greater describes a broad distribution (Cook et al., 2014). As well, 12 linear regression analyses were conducted to investigate if the overall magnitude of kinematic changes (RMSD of all joints) in each task (four tasks) had a linear relationship with power output, change in power, or change in jump height.

Research Question #4

Hypothesis 4.1 investigated the effect of age, trunk flexion angle at peak GRF, and trunk flexion angle at peak power on specific lower-body power output during CMJ and was tested via a single multiple linear regression analysis. A single factor ANOVA was conducted to investigate the effect of age (four levels) on trunk flexion angle at peak GRF. A second single factor

ANOVA was conducted to investigate the effect of age (four levels) on trunk flexion angle at the instant of peak power production. Hypothesis 4.2 investigated the effect of age, hip flexion angle at peak GRF, and hip flexion angle at peak power on specific lower-body power output during CMJ via a single multiple linear regression analysis. A single factor ANOVA was conducted to investigate the effect of age (four levels) on hip flexion angle at peak GRF. A second single factor ANOVA was conducted to investigate the effect of age (four levels) on hip flexion angle at the instant of peak power production.

Research Question #5

For Hypothesis 5.1, a linear regression analysis was performed to investigate if plantar flexor FA generated a statistically significant predictive model of ankle power. For Hypothesis 5.2, 16 95% CIs based on the t-distribution were used to determine if there were significant differences in FA of four prime movers at the ankle across four athletic groups (short-distance, long-distance, non-athlete and overall). For Hypothesis 5.3, a single linear regression model was developed to investigate whether muscle fiber proportion (taken from literature) could generate a statistically significant predictive model of FA.

“We do not stop exercising because we grow old – we grow old because we stop exercising”

- Dr. Kenneth H Cooper

4 Research Question #1: What is the effect of age, sex and activity level on lower-body specific power, and specific joint power during all functional tasks?

Introduction

The loss of muscle mass with age (sarcopenia), can result in losses of approximately 50% by the eighth to ninth decade of life (Wilkinson et al., 2018). Similarly, losses in muscular strength (dynapenia), begin around fourth or fifth decade and can result in lower body strength losses of approximately 50-60% by the eighth decade (Bassey et al., 1992; Hunter et al., 2016). Lastly, muscular power which is strongly implicated in physical function (Foldvari et al., 2000), can be reduced by approximately 75% between the age of 20-31, and 74-91 (Bassey and Short, 1990). Fortunately, losses in muscular power can be mitigated to some extent by general physical activity (PA; Foldvari et al., 2000; Ramsey et al., 2021) and exercise (Balachandran et al., 2022; Byrne et al., 2016; McPhee et al., 2016). This research question sought to understand the effect of age and PA on lower-body power during a maximal countermovement jump (CMJ). This research also sought to understand how these factors might affect joint power during the CMJ, and the downstream implications on joint power, if any, on several functional tasks.

Hypotheses

1.1 Lower-body power during the CMJ would decrease with age but increase with activity level.

1.2 Joint power during the CMJ and functional tasks would decrease with age but increase with activity level.

Methods

The sample size for Hypothesis 1.1 was 62 (39 males and 23 females) consisting of athletes and non-athletes as reported in Common Methods. The average sample size for Hypothesis 1.2 was 57, with stair descent hip power having the lowest sample size of 47 due to motion capture volume complications. As well, one participant’s data (056) was omitted from the stair joint power analysis (Hypothesis 1.2) due to outlier joint powers which likely resulted from inconsistent amplifier gain leading to erroneous GRFs in the one stair force plate. Protocol for

Hypothesis 1.1 and 1.2 consisted of CMJs, chair rises, stair ascents and stair descents performed with motion capture and force plates facilitating the calculation of lower body power via net force and pelvis CoM velocity, and inverse dynamics for joint power calculation. Activity level was determined with multiple questionnaires however the HUNT survey was applied here. Analyses for Hypotheses 1.1 and 1.2 involved multiple linear regression. The additional findings related to normalization and body fat relied on linear regression and univariate ANOVA. More detailed methods for this research question can be found in common methods section.

Results

Multiple linear regression was used to determine the extent to which age, sex and activity level predicted lower-body specific power during the CMJ (Hypothesis 1.1). The strength of this model appeared dependent on the method used to quantify PA. Several continuous PA variables were evaluated for their strength of contribution to the model. These variables were IPAQ MET minutes per week, IPAQ Vigorous MET minutes per week, and HUNT survey score.

In the first model, age, sex and PA via IPAQ MET minutes generated a statistically significant predictive model of lower-body specific power during the CMJ ($F_{3,59}$ = 18.975, p<0.001, R^2=0.504). However, only age and sex made statistically significant contributions to the model (p<0.022). The contribution of activity level, quantified as weekly METs of activity from the IPAQ survey, did not make a statistically significant contribution to the model (p=0.183).

In the second model, age, sex and PA via IPAQ vigorous minutes generated a statistically significant predictive model of lower-body specific power during the CMJ ($F_{3,59}$ = 20.357, p<0.001, R^2=0.522). However, once again, only age and sex made statistically significant contributions to the model (p<0.022). The contribution of activity level, quantified as weekly METs of vigorous PA from the IPAQ survey, did not make a statistically significant contribution to the model (p=0.052).

In the third model (**Figure 4.1**), age, sex and PA via HUNT survey generated the strongest statistically significant predictive model of lower-body specific power during the CMJ ($F_{3,62}$ = 29.960, p<0.001, R^2=0.608), with all independent variables making statistically significant contributions to the model (p<0.01; **Table 4.1**). No significant interaction effects between age and activity (p=0.108), between age and sex (p=0.698) or between sex and activity level (p=0.690) were observed. The HUNT survey was used as the activity level metric for all remaining activity level analyses.

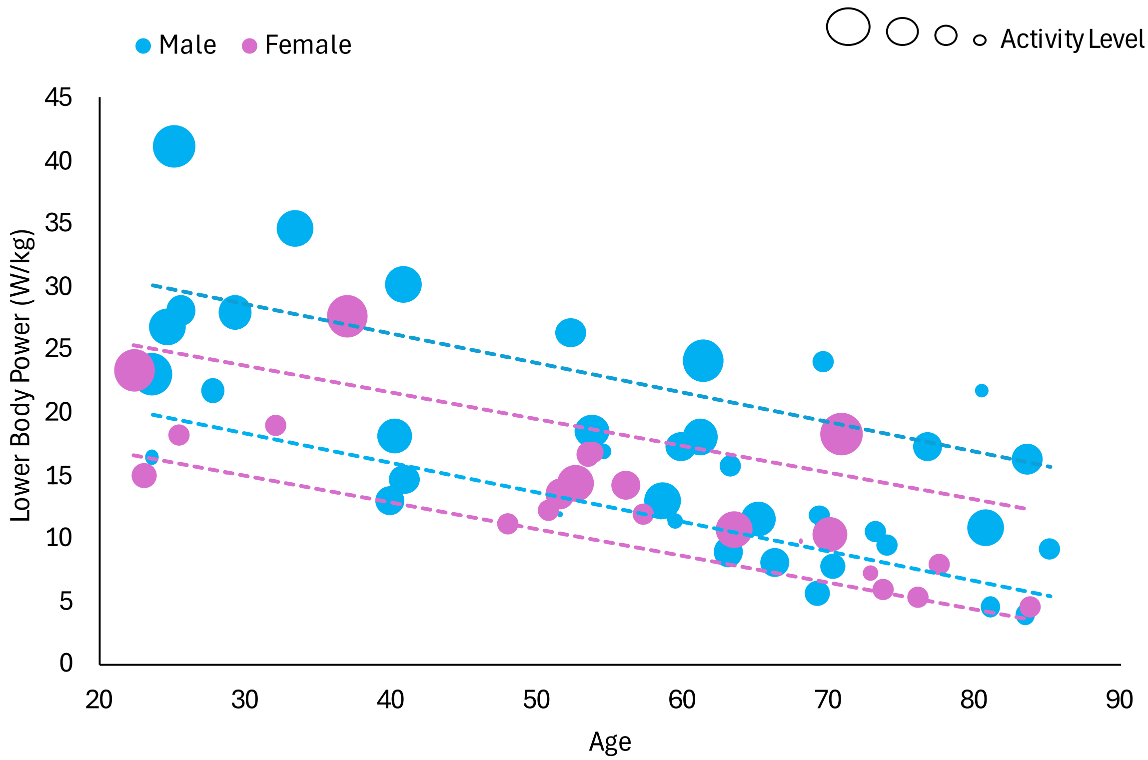

Figure 4.1 – The effect of age, sex and activity level on peak lower-body power. The linear regression model ($F_{3,62}$ = 29.960, p<0.001, Intercept = 25.305, R^2=0.608) predicted annualized reductions in power of 0.234W/kg per year in males and 0.212W/kg per year in females. Trendlines represent the predicted power outputs across age for the highest (score of 15, top line) and lowest (score of 0, bottom line) possible physical activity levels. Male and female are shown by separate color and bubble size shows activity level. By following a trendline, one can compare the predicted power output of a 30-year-old inactive female to the power output of an 80-year-old inactive (or active) female. Likewise, by comparing the gap between trendlines, the effect of physical activity on power output can be observed directly.

Table 4.1 – The linear regression models for lower-body power during the CMJ. Age, sex and activity level generated a statistically significant model for lower-body specific power during the countermovement jump with all independent variables making statistically significant contributions to the model. No significant age*HUNT interaction was found for any model below (p>0.077). Correlations presented are zero-order.

CMJ					
DVs	**IVs: age, sex and activity level (HUNT)**				
Lower-body power	**Model**	$F_{3,61}$ = 29.960, **p<0.001**, Intercept = 25.305, R^2=0.608			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.228	0.035	**<0.001**	-0.654
	Sex	-3.490	1.319	**0.010**	-0.221
	Activity Level	0.648	0.159	**<0.001**	0.540
Lower-body power, male	**Model**	$F_{2,38}$ = 20.933, **p<0.001,** Intercept = 25.364, R^2=0.512			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.234	0.054	**<0.001**	-0.671
	Activity Level	0.688	0.264	**0.013**	0.545
Lower-body power, female	**Model**	$F_{2,22}$ = 36.778, **p<0.001,** Intercept = 21.333, R^2=0.765			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.212	0.032	**<0.001**	-0.755
	Activity Level	0.587	0.131	**<0.001**	0.559

Males | **Females**

a) Predicted power (W/kg)

Males

	Activity Level (HUNT)				
	0	**3.75**	**7.5**	**10**	**15**
Age 30	18.3	20.9	23.5	25.2	28.7
Age 80	6.6	9.2	11.8	13.5	17.0

Females

	Activity Level (HUNT)				
	0	**3.75**	**7.5**	**10**	**15**
Age 30	15.0	17.2	19.4	20.8	23.8
Age 80	4.4	6.6	8.8	10.2	13.2

b) Predicted power difference over 50 years (W/kg)

Males

	Age 80				
Age 30	**0**	**3.75**	**7.5**	**10**	**15**
0	-11.7	-9.1	-6.5	-4.8	-1.4
3.75	-14.3	-11.7	-9.1	-7.4	-4.0
7.5	-16.9	-14.3	-11.7	-10.0	-6.5
10	-18.6	-16.0	-13.4	-11.7	-8.3
15	-22.0	-19.4	-16.9	-15.1	-11.7

Females

	Age 80				
Age 30	**0**	**3.75**	**7.5**	**10**	**15**
0	-10.6	-8.4	-6.2	-4.7	-1.8
3.75	-12.8	-10.6	-8.4	-6.9	-4.0
7.5	-15.0	-12.8	-10.6	-9.1	-6.2
10	-16.5	-14.3	-12.1	-10.6	-7.7
15	-19.4	-17.2	-15.0	-13.5	-10.6

c) Predicted power difference (%)

Males

	Age 80				
Age 30	**0**	**3.75**	**7.5**	**10**	**15**
0	-63.8%	-49.7%	-35.7%	-26.3%	-7.5%
3.75	-68.2%	-55.9%	-43.6%	-35.4%	-18.9%
7.5	-71.7%	-60.8%	-49.8%	-42.5%	-27.8%
10	-73.7%	-63.4%	-53.2%	-46.4%	-32.7%
15	-76.8%	-67.8%	-58.8%	-52.8%	-40.8%

Females

	Age 80				
Age 30	**0**	**3.75**	**7.5**	**10**	**15**
0	-70.8%	-56.1%	-41.4%	-31.6%	-12.0%
3.75	-74.5%	-61.7%	-48.9%	-40.4%	-23.3%
7.5	-77.4%	-66.1%	-54.7%	-47.1%	-32.0%
10	-79.0%	-68.5%	-57.9%	-50.9%	-36.8%
15	-81.6%	-72.4%	-63.1%	-56.9%	-44.6%

d) Per annum predicted power difference (% p.a.)

Males

	Age 80				
Age 30	**0**	**3.75**	**7.5**	**10**	**15**
0	-1.28%	-0.99%	-0.71%	-0.53%	-0.15%
3.75	-1.36%	-1.12%	-0.87%	-0.71%	-0.38%
7.5	-1.43%	-1.22%	-1.00%	-0.85%	-0.56%
10	-1.47%	-1.27%	-1.06%	-0.93%	-0.65%
15	-1.54%	-1.36%	-1.18%	-1.06%	-0.82%

Females

	Age 80				
Age 30	**0**	**3.75**	**7.5**	**10**	**15**
0	-1.42%	-1.12%	-0.83%	-0.63%	-0.24%
3.75	-1.49%	-1.23%	-0.98%	-0.81%	-0.47%
7.5	-1.55%	-1.32%	-1.09%	-0.94%	-0.64%
10	-1.58%	-1.37%	-1.16%	-1.02%	-0.74%
15	-1.63%	-1.45%	-1.26%	-1.14%	-0.89%

Figure 4.2 – A tabular summary of predictions from the regression model. Activity level was a strong moderating factor in the regression analysis. As shown in part (a), the predicted power output for a highly active 80-year-old male (17.0W/kg) is close to the predicted power output of an inactive 30-year-old male (18.3W/kg). A similar relationship was found for highly active 80-year-old females (13.2W/kg) compared to inactive 30-year-old females (15.0W/kg). Part (b) shows the differences in power between young and old adults for various activity levels. Crucially, although younger males begin with greater power, older males are predicted to lose more than females even with activity level held constant (-22W/kg for males versus -19.4W/kg for females at zero activity, for example). Part (c) shows the predicted percentage difference in power for old relative to young participants varying with activity level. Part (d) shows the same predicted percentage difference in power divided by the 50 years between age groups, providing an annualized rate of power loss. Consistent throughout is the concept that older adults with high activity levels demonstrate greater retention of power output.

Figures 4.1 and 4.2 demonstrate several important points regarding the impact of PA on power output in the CMJ. Firstly, **Figure 4.1** shows that the rate of power loss in males was greater than that of females (0.234W/kg versus 0.212W/kg per year), however males demonstrated 3.5W/kg greater power on average across the age-span and thus are likely more tolerant to a greater loss rate. **Figure 4.2** demonstrates example cases of predictions of power output for a young adult (30 years) and an old adult (80 years) with varying activity levels. Part (a) predicts that a highly active older adult could maintain power output like that of an inactive younger adult (17W/kg versus 18.3W/kg), which is a difference of only 1.4W/kg or 7.5%. Thus, the highly active older adult can expect to lose 0.15% power annually from age 30 to age 80. However, this figure is with reference to an inactive 30-year-old male. If instead activity level is held constant at the male average of 7.05 in this cohort, from age 30 to age 80, this cross-sectional model predicts an annual loss of 1.01% power. A similar result is predicted for a highly active older female versus an inactive younger female (13.2W/kg versus 15W/kg) which is a difference of 1.8W/kg or 12.0%. Once again, if activity level is held constant at the female average of 6.19, from age 30 to age 80, this cross-sectional model predicts an annual loss of 1.14% power. Of note, when expressed in absolute terms (i.e. W/kg per year), males appear to lose power at a greater rate however when expressed as a percentage as done above, females show greater predicted loss rates due to lower initial power and reduced response to PA as is evident in the PA regression coefficients presented in **Table 4.1** above. The male PA coefficient was 0.687 whereas the female coefficient was 0.588W/kg per year

An important consideration is that these data are cross-sectional and are not a result of following individuals over the course of 50 years. Thus, discussion of “power loss” does not actually describe power that was lost and measured in one individual over time but instead refers to the lower power observed across the age-span in this cohort.

The effect of age, sex and activity level was also explored in the CMJ at the joint level (Hypothesis 1.2). With respect to *specific ankle power* during the CMJ, age, sex and physical level generated a statistically significant predictive model ($F_{3,59}$ = 21.416, p<0.001, R^2=0.534), however only age and sex made statistically significant contributions to the model (p<0.025). With respect to *specific knee power* during the CMJ, age, sex and PA level generated a statistically significant predictive model ($F_{3,59}$ = 11.974, p<0.001, R^2=0.391) however only age and activity level made statistically significant contributions to the model (p<0.016). With respect to *specific hip power* during the CMJ, age, sex and PA level generated a statistically significant predictive model ($F_{3,58}$ = 12.997, p<0.001, R^2=0.415) however only age and activity level made statistically significant contributions to the model (p<0.002). A tabular summary of the CMJ joint-level regression model is provided in **Table 4.2**.

Table 4.2 – The linear regression models for joint power assessment during the CMJ. For joint-level regression analyses, only age made consistent, statistically significant contribution to the model. Correlations presented are zeroth order.

CMJ					
DVs	**IVs: age, sex and activity level (HUNT)**				
Ankle power	**Model**	$F_{3,59}$ = 21.416, **p<0.001**, R^2=0.534, Intercept = 11.799			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.078	0.012	**<0.001**	-0.671
	Sex	-1.034	0.448	**0.025**	-0.170
	Activity Level	0.103	0.053	0.057	0.385
Knee power	**Model**	$F_{3,59}$ = 11.974, **p<0.001**, R^2=0.391, Intercept = 9.020			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.069	0.016	**<0.001**	-0.530
	Sex	-1.056	0.619	0.094	-0.158
	Activity Level	0.182	0.073	**0.016**	0.421
Hip power	**Model**	$F_{3,59}$ = 12.997, **p<0.001**, R^2=0.415, Intercept = 6.469			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.050	0.012	**<0.001**	-0.516
	Sex	-0.571	0.485	0.244	-0.139
	Activity Level	0.193	0.058	**0.002**	0.491

Regression models were also generated for each joint (ankle, knee and hip) during the three functional tasks (stair ascent, stair descent and chair rise). The findings for these analyses are summarized in **Table 4.3, 4.4 & 4.5** below. Briefly, hip power during the chair rise task generated a statistically significant predictive model ($F_{3,59}$ = 2.924, p<0.042, R^2=0.135) with only PA level making statistically significant contributions to the model. No other joint during any task generated a statistically significant model (p>0.087).

Table 4.3 – The linear regression models for each joint during the chair rise task. Age, sex and activity level generated a statistically significant model hip power during the chair rise task (p=0.042, R^2 = 0.135) with only activity level making a statistically significant contribution to the model (p=0.006, r = 0.356). Age, sex and activity level did not generate statistically significant models for either ankle or knee power (p>0.209). Correlations presented are zero-order.

Chair Rise					
DVs	**IVs: age, sex and activity level (continuous HUNT)**				
Ankle power	**Model**	$F_{3,59}$ = 0.453, p=0.716, R^2 = 0.024			
Knee power	**Model**	$F_{3,59}$ = 1.563, p=0.209, R^2 = 0.077			
Hip power	**Model**	$F_{3,59}$ = 2.924, **p=0.042**, R^2 = 0.135			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	0.002	0.003	0.534	-0.022
	Sex	-0.045	0.123	0.717	-0.081
	Activity Level	0.042	0.015	**0.006**	0.356

Table 4.4 – The linear regression models for each joint during the stair ascent task. Age, sex and activity level did not generate a statistically significant model for either ankle, knee or hip power during the stair ascent task (p>0.079).

Stair Ascent		
DVs	**IVs: age, sex and activity level (continuous HUNT)**	
Ankle power	**Model**	$F_{3,55}$ = 1.250, p=0.301, R^2= 0.013
Knee power	**Model**	$F_{3,55}$ = 2.255, p=0.093, R^2= 0.115
Hip power	**Model**	$F_{3,54}$ = 2.400, p=0.079, R^2= 0.072

Table 4.5 – The linear regression models for each joint during the stair descent task. Age, sex and activity level did not generate a statistically significant model for either ankle, knee or hip power during the stair ascent task (p>0.239).

Stair Descent		
DVs	**IVs: age, sex and activity level (continuous HUNT)**	
Ankle power	**Model**	$F_{3,55}$ = 1.450, p=0.239, R^2 = 0.024
Knee power	**Model**	$F_{3,54}$ = 0.599, p=0.618, R^2 = 0.023
Hip power	**Model**	$F_{3,46}$ = 0.651, p=0.587, R^2 = 0.023

Key Findings

- 0.234W/kg of lower body specific power per year lost in males, 0.212W/kg in females
- 0.688W/kg more lower body power for every unit increase in PA in males, 0.587W/kg in females
- Ankle, knee and hip joint power during CMJ decreased with age
  - Knee and hip joint power increased with PA
- Bodyfat was significantly greater in underactive adults over the age of 50 years (p<0.011), which may affect interpretation of age-related power loss.

Additional Findings

The HUNT survey allows for categorical scoring as active or underactive when interpreted in accordance with the WHO activity guidelines for adults (including older adults aged 65+) (Bull et al., 2020; Lerfald et al., 2023). Comparing means of active versus underactive adults via ANOVA can also be useful for generalizing the findings. A three-way ANOVA was used to analyze specific joint power during the CMJ as a function of age (<30, 30-50, 50-70, >70), sex (male or female) and activity via HUNT survey (active or underactive). The analysis revealed no significant interaction effects ($p>0.169$, $\eta_p^2 < 0.101$) but did uncover significant main effects of age ($F_{3,61} = 5.785$, $p=0.002$, $\eta_p^2 = 0.270$; **Figure 4.3**) and activity level ($F_{1,61} = 14.849$, $p<0.001$, $\eta_p^2 = 0.240$; **Figure 4.4**). Post-hoc testing on age groups with Bonferroni correction determined that the <30 age-group demonstrated significantly greater specific power output than >70 age-group (Δ = 8.95W/kg, p=0.004). As well, the 30-50 age-group had greater specific power output compared to the 50-70 age group (Δ = 7.627W/kg, p=0.033) and the >70 age-group (Δ = 10.473W/kg, p=0.003).

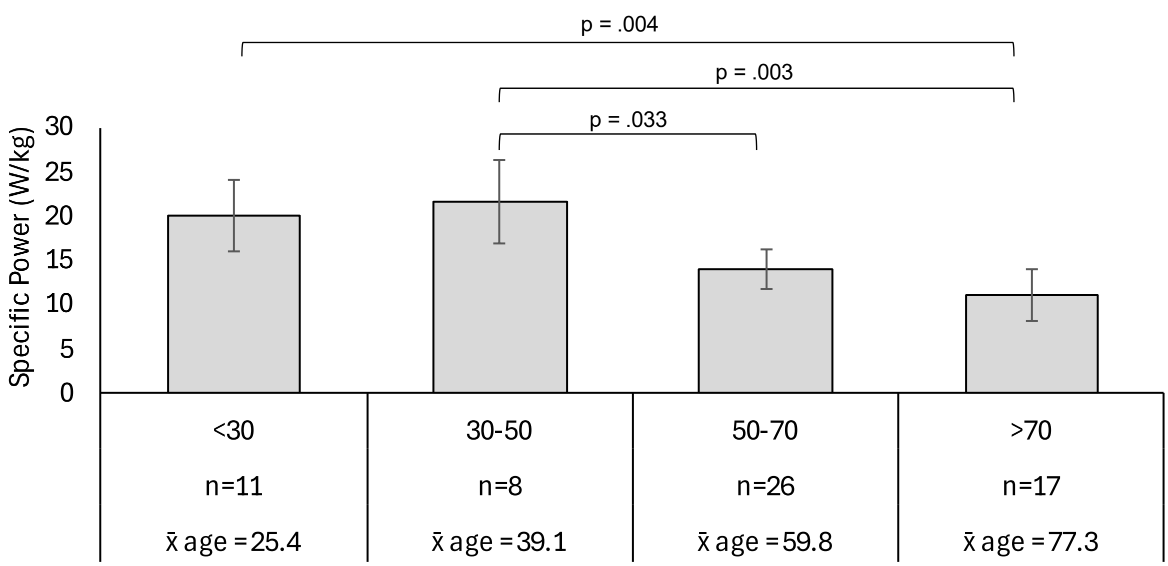

Figure 4.3 – The effect of age on specific lower body power. A three-factor ANOVA showed that those aged less than 30 displayed greater lower-body specific power in CMJ compared to those aged greater than 70 years (Δ = 8.95W/kg, p<0.004). As well, 30–50-year-olds displayed significantly greater power than those aged 50-70 years of age (Δ = 7.627W/kg, p=0.033) and greater than 70-years (Δ = 10.473W/kg, p=0.003).

With respect to activity level, the underactive group showed significantly less specific power during the CMJ compared to the active group (Δ = 7.211W/kg, p<0.001), however the active group was also 11.5 years younger than the underactive group and therefore this difference should be interpreted cautiously (**Figure 4.4**). Per the regression analysis conducted prior (**Table 4.1**), in this cohort, increasing age by one year corresponded to a decreased power of 0.228W/kg. Therefore, an 11.5 year age difference corresponds to a difference in power of 2.622W/kg which may be sufficient to nullify the statistically significant difference between active and underactive participants.

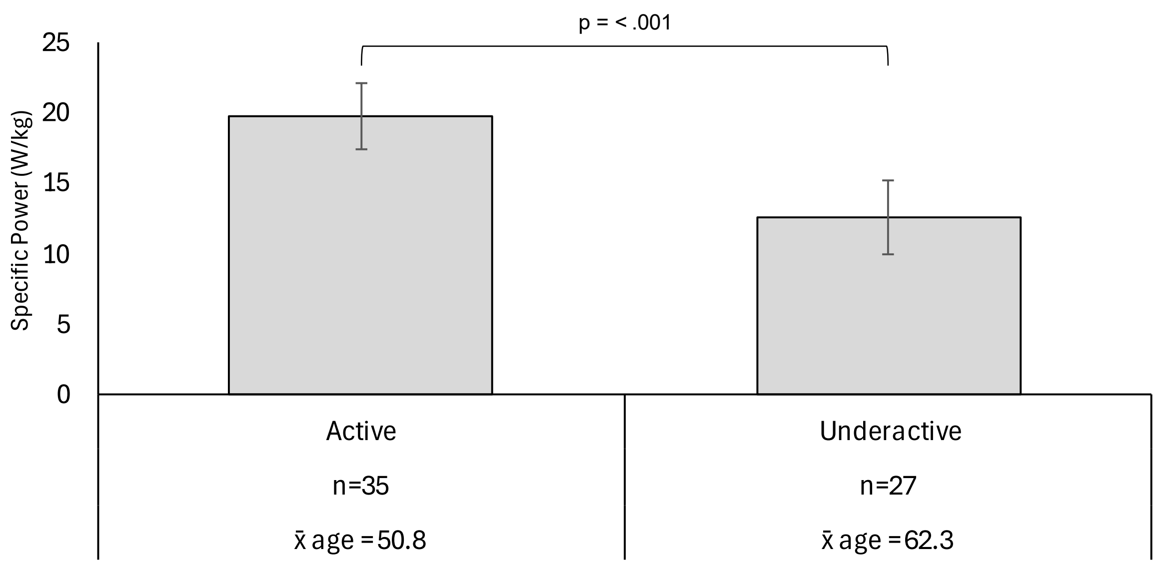

Figure 4.4 – The effect of activity level on lower-body power. A three-factor ANOVA determined that active participants displayed greater specific power output compared to underactive individuals (Δ = 7.211W/kg, $p<0.001$). Active individuals were determined to meet the 2020 WHO guidelines for physical activity.

Four separate three-way MANOVAs were used to analyze specific joint power in each task (CMJ, chair, ascent, descent). Each MANOVA assessed the effect of age (<30, 30-50, 50-70, 70+), sex (male or female) and activity via HUNT survey (active or underactive) on peak specific joint power (ankle, knee, and hip). For the CMJ, no interactions were detected ($p>0.112$) but a significant main effect for age and activity level were found for ankle, knee and hip joints (6 main effects – **Table 4.6**).

Table 4.6 – A tabular summary of MANOVA findings on joint-power during CMJ. A three-factor MANOVA evaluated the effect of age, sex and activity level on the four dependent variables lower-body power, ankle power, knee power, and hip power. No interaction effects were observed however, a main effect for age and activity was observed for lower-body power in addition to all three joint powers (p<0.039).

DVs	**MANOVA for CMJ power**				
	IV	**F**	**df_{group}**	**df_{total}**	
Ankle power	**Age**	5.822	3	59	**0.002**
	Sex	0.291	1	59	0.592
	Activity	12.160	1	59	**0.001**
Knee power	**Age**	3.032	3	59	**0.039**
	Sex	0.810	1	59	0.373
	Activity	5.021	1	59	**0.030**
Hip power	**Age**	3.674	3	59	**0.019**
	Sex	0.094	1	59	0.760
	Activity	5.085	1	59	**0.020**

A three-way MANOVA investigating the *chair rise task* failed to detect a statistically significant interaction or main effect of age, sex or activity classification on power at any of the three lower-limb joints (p>0.108). An additional three-way MANOVA investigating the *stair ascent task* failed to detect a significant interaction or main effect of age, sex or activity classification on power at any of the three lower-limb joints (p>0.069). A final three-way MANOVA investigating the *stair descent task* failed to detect a significant interaction or main effect of age, sex or activity classification on power at any of the three lower-limb joints (p>0.137). These data suggest that, in contrast to the CMJ, age and activity level do not affect joint power output during the submaximal functional tasks (chair, stair ascent and descent).

Thus far, all power data presented have all been normalized to body mass. This is conventional practice as it allows for comparison across individuals of differing body mass, and is also relevant in the sense that activities of daily living typically involve weight-bearing locomotion (Alcazar et al., 2021). However, there also exists some concern in the literature as to whether this normalization method is fair for older adults since they typically have greater fat mass (Alvero-Cruz et al., 2021). Instead of normalizing power to body mass, power can be normalized to muscle volume or muscle cross-sectional area instead, alleviating the concerns related to greater fat mass in older adults. It also may give a more precise measure of a specific joint's ability to generate or absorb power based on the amount of muscle mass at that specific joint. To investigate the extent of this issue in this cohort, a multiple linear regression for the effect of age on knee power was evaluated using two variations of normalized knee power. The first was the standard knee power (Watts) divided by body mass (kg) – the analysis for this was conducted above and was summarized in **Table 4.6**. The second normalization method involved knee power (Watts) divided by CSA of the knee extensor musculature (cm^2) measured by MRI in a subsample of this cohort. Age, sex and activity level failed to generate a statistically significant predictive model of knee power normalized to CSA of the knee extensors ($F_{3,35}$ = 0.209, p=0.890, R^2 = 0.019; **Figure 4.5**).

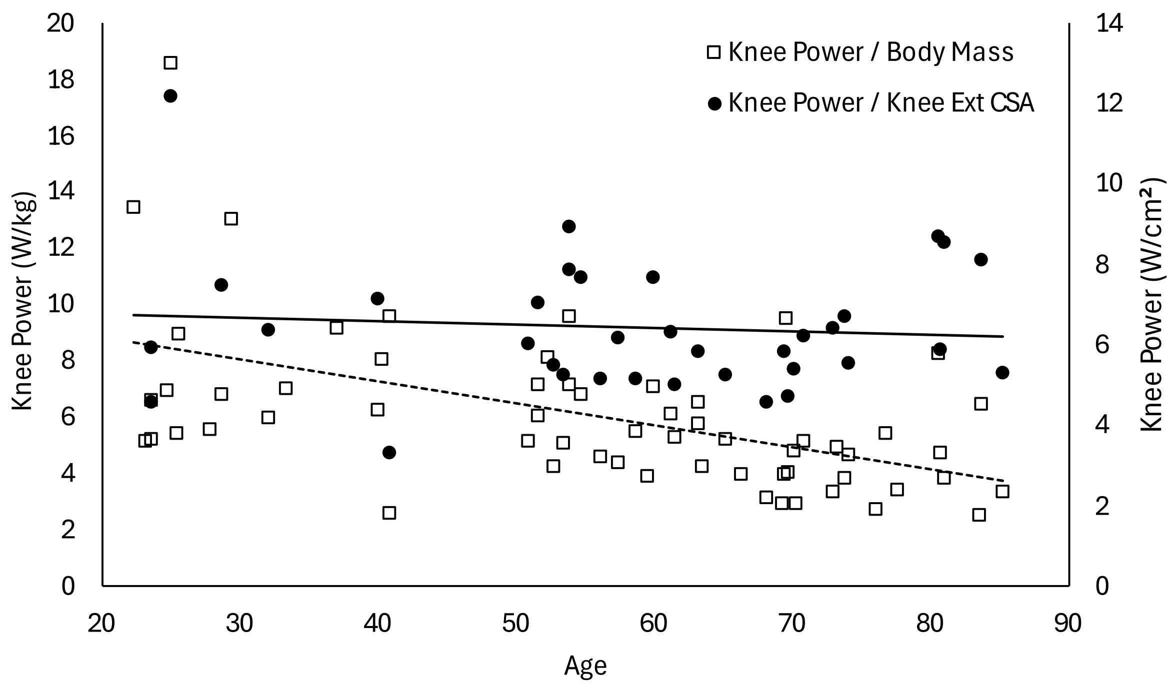

Figure 4.5 – Linear regression models based on knee power during the CMJ, normalized to body mass (successful model) and to CSA of the knee extensor muscles (unsuccessful model).

Linear regression failed to show a statistically significant linear relationship between power normalized to knee extensor CSA and age. Considering that sample size was nearly halved due to low MRI participation, it is possible that this analysis was underpowered. Nonetheless, it does appear that the slope would be lower than that of power normalized to body mass implying that rates of decline in power are sensitive to normalization methods.

Subsequent analysis via ANOVA into the effect of age, sex and activity level on bodyfat attempted to explain the null finding above. Testing revealed a significant interaction effect for age and activity classification ($F_{3,56}$ = 3.172, p=0.034) on bodyfat levels. As shown below in **Figure 4.6**, active participants aged 50-70 had significantly less bodyfat compared to their age-matched underactive counterparts (Δ = 6.1%, p=0.011) and likewise for participants aged older than 70 (Δ = 15.0%, p<0.001). The analysis also revealed main effects of age ($F_{3,56}$ = 3.560, p=0.020), sex ($F_{2,56}$ = 11.964, p=0.001) and activity level ($F_{1,56}$=15.712, p<0.001) on bodyfat percentage.

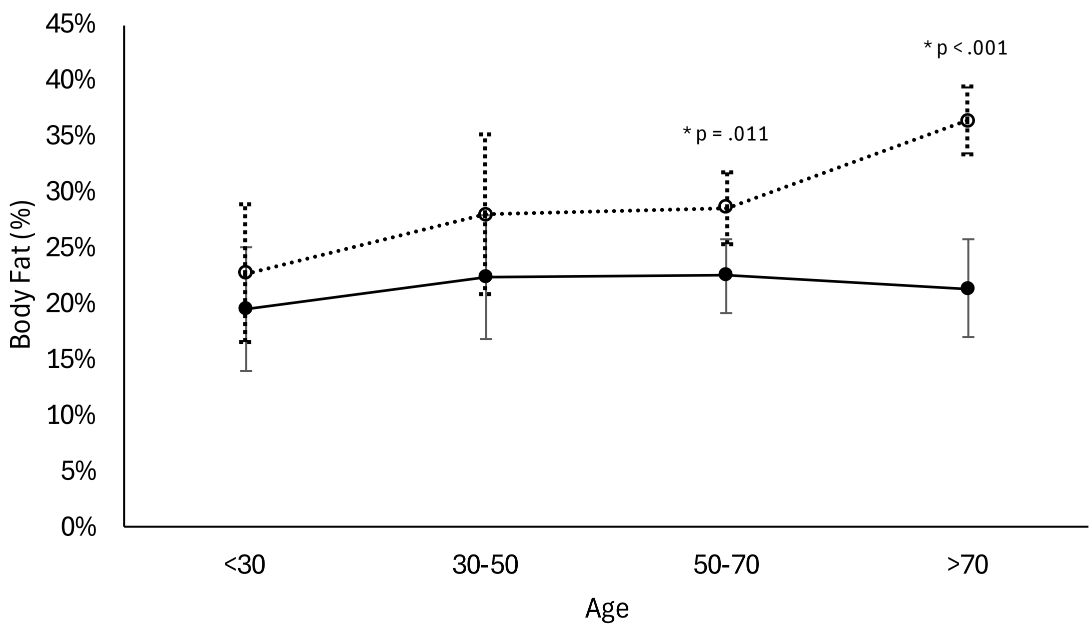

Figure 4.6 – Bodyfat percentages by age group, separated by active and underactive groups. Percentage bodyfat was significantly greater in underactive older adults over the age of 50. The effect was larger by age 70 with underactive older adults having 15.0% greater bodyfat percentage. Solid line shows the active group, and the dashed line shows the underactive group.

This suggests that power output normalized to body mass (**Figure 4.5** above, open squares) may underestimate the performance of older adults who are under-active due to higher body fat levels. Older adults who remain active may not be affected as their body fat levels did not differ significantly at any age group in this cohort.

Discussion

Overview

This section will review findings related to absolute and yearly percentage rates of power loss, compares them to existing literature, and offers some likely reasons for discrepancies are offered. The effects of sex and activity level are discussed as well – the effect of sex appears inconsistent and while the effect of activity level is always protective, the particular strength of the relationship depends on the means of measuring PA level.

Annual Absolute Rate of Power Loss

Linear regression analysis found that specific power during the CMJ decreased at a rate of 0.228W/kg per year or 0.234W/kg in males and 0.212W/kg in females. Thus, in absolute terms, males were projected to lose lower body power at a greater rate than females. In 2004, Runge and colleagues measured power output as a function of age in 169 women and 89 men aged 18 to 88 years. They found a per annum loss rate of 0.62W/kg in males and 0.42W/kg in females. The steeper loss rate in males compared to females is consistent with the findings presented here however the slopes were greater in previous research. Part of this discrepancy may result from arm swing which was allowed by Runge and colleagues. Arm swing is known to increase power output in the vertical jump by to 16% (Shetty and Etnyre, 1989) to 21% (Harman, Everett et al., 1990). Secondly, the Leonardo Jump Mechanography apparatus computes power internally by taking the product of integrated acceleration (velocity) and net force. This differs from the power calculation used in this research which evaluated the product of vertical velocity (derived position data, filtered at 10Hz) and net ground reaction force (also filtered at 10Hz). Runge and colleagues estimated power losses of approximately 55% between the ages of 20 and 80 which is strikingly consistent with the results presented here which show losses of 58% between the same ages. Note that **Figure 4.2** shows the differences between 30 and 80 years of age – the figures for 20 years of age can be calculated via regression equations presented in **Table 4.1** however that is an extrapolation of this dataset as we did not measure power in those aged less than 22.4 years (and is why **Figure 4.2** uses 30 years as the young reference). In 2015, Siglinski and colleagues evaluated various physical capacities in 332 participants of mean age 65.7, ranging from 27 to 96 years of age. Like the data presented here, their linear regression model found a steeper drop-off in specific power in males versus females (0.458W/kg per year in males versus 0.280Wkg per year in females). Both their male and female data indicated accelerated losses in power (steeper slopes) likely arising from several very high power outputs observed in their younger participants (one in excess of 60W/kg, whereas 41.1W/kg was the highest in this data). Siglinksi and colleagues do not clarify whether their participants were allowed the free use of arms which would result in greater power output as discussed above. As well, their assessment also used a Leonardo Jump Mechanography platform which processes jump data internally as mentioned above. How exactly this analysis differs this research is currently unclear and thus a direct comparison of slopes is difficult. Nonetheless, in summary, the data presented is consistent with the work of Runge and colleagues, and Siglinsky and colleagues in that males lose specific power at a greater rate than females and that an individual can expect to lose more than 50% of their lower body power through their lifespan. The specific per annum loss rate (W/kg per year) depends on the type of CMJ, and on the technology used to measure power output.

Annual Percentage Rate of Power Loss

When expressed as a percentage, calculated relative to the predicted power of a 30-year-old performing the average activity level (7.05 in males, 6.19 in females), contrary to the above, the per annum loss rates were greater in females (1.14% per year) than in males (1.01% per year). This is a result of lower predicted power in females at the 30-year reference age, and the lower predicted response to PA in females – males were predicted to gain 0.688W/kg for every unit increase in HUNT survey score, whereas females were predicted to gain 0.587W/kg for each unit increase. This difference in sensitivity to activity level predicts that a male would have an additional 1.5W/kg of lower body power at maximum activity level compared to a female of the same age and activity level. In 1994, Skelton and colleagues found a per annum rate of power loss of 3.0% in males and 1.7% in females. This rate is three times greater than the rate presented here for this study's male cohort. The per annum calculation used by Skelton and colleagues is not clearly detailed but appears to be based on an age range of 65 to 85 years. Thus, their per annum loss rates are expected to be higher if they are expressed relative to a greater reference age (65 years versus the 30-year reference age used here) due to the lower power at that age. If per annum power loss rates are calculated using the present data over a 20-year span from 65 to 85, with a 65-year-old reference age as Skelton and colleagues appear to have done, the rate of power loss in our cohort rises to 1.56% per year in males and 1.90% in females. In doing so, our power loss rates begin to approach those of Skelton and colleagues. As well, our cohort included 30 athletes and 32 non-athletes. The implication is that our regression model is based on higher-than-normal activity levels at all ages, due to the participation of athletes. In fact, 95% confidence intervals for activity level in athletes was 6.98 to 9.83, and in non-athletes it was 3.77 to 6.53. Thus, differences in cohort may also help explain the lower per annum power loss in our study. Of note, Skelton and colleagues did report absolute power loss rates (W/kg per year) however a direct comparison is probably not appropriate for the following reasons. Firstly, the Nottingham power rig used in their study (a lower body ergometer) measures power output from a single limb and thus is not surprising that the absolute power outputs were 6-fold less than ours. As well, the Nottingham Power Rig is typically configured to complete the repetition prior to complete leg extension to minimize discomfort and risk related to hyperextension (Bassey and Short, 1990). As such, it does not permit true ballistic movement like the CMJ and thus would be expected to show lesser power. Though, the Nottingham Power Rig does show strong agreement with force plate jumps (ρ=0.86, $p<0.001$). Nevertheless, with these factors in consideration, a direct comparison of absolute or specific power across the age-span was not appropriate, and while imperfect, the per annum percentage figures appear to be the most suitable way to draw comparison between our work and that of Skelton and colleagues.

Therefore, in summary, our per annum percentage loss rates of 1.01% per year in males and 1.56% per year in females are lower than previously reported by Skelton and colleagues, however this discrepancy can likely be explained by our per annum rates being calculated relative to a younger reference age and by our cohort including many highly active athletes.

Sex Differences

The initial regression model found that sex contributed significantly to the analysis. In 2015, Siglinksi and colleagues investigated jump power and jump height using a Leonardo Jump Mechanography platform in 213 females and 119 males with average age of 65.4 for the whole cohort. Siglinski and colleagues reported a significant difference in jump power between males and females (28.5 vs 21.9W/kg, $p<0.001$). Alvero-Cruz and colleagues (2021) tested power output in 162 male and 91 female masters athletes from power, endurance and mixed athletic disciplines (median age of 58 and 55 for males and females). They found significant differences in peak power between males and females (38.8W/kg vs 32.91W/kg) despite comparable age. However, while the actual numbers were not published, by graphical estimation it would appear that a great majority of the power athletes were male and thus the male-female comparison also contains a comparison of athletic discipline. Our study suffers the same flaw with only one female sprinter. It appears to be a natural consequence of the low female short distance athlete population. As well, Alvero-Cruz and colleagues allowed participants to swing their arms which can increase power by approximately 20%. As for contrasting evidence, In 1990, Bassey and Short used the Nottingham Power Rig to assess lower body power in a 20 to 93-year-old age-group. They found that large decrements in power occurred for both sexes, and the rate of loss was similar between sexes when normalized to body mass (Bassey and Short, 1990). The authors noted as well that there was no increase in body mass with age for males or females and that changes to body mass could not explain the changes in specific power. As an aside, our additional findings related to fat mass paint a slightly different picture. In our data, an interaction effect was discovered between age and activity level on body fat percentage. Participants over the age of 50 who were classified as underactive had significantly greater body fat than their active age-matched peers. This implies that underactive adults aged greater than 50 may have higher amounts of body fat and thus will display lower specific powers simply by virtue of having more fat mass and less contractile mass in the denominator used during normalization (body mass). This may result in underestimated specific power in that particular group. In 2018, Piasecki and colleagues investigated CMJ power in 38 sprinters, 149 long-distance runners and 59 non-athlete controls aged greater that 60 years (approximate mean age of 70 years). They found no significant difference in power by sex, but did find significant differences between athletic disciplines, which will be discussed later in Research Question #2. In 2016, Glenn and colleagues investigated the effect of sex and athletic discipline on power output during a maximal sit-to-stand task in adults aged approximately 60 years, including a recreationally active control group. Participants were in structed to stand as quickly as possible while their power was assessed via a Tendo unit (a linear position transducer). They found a significant effect of activity level on peak power but no sex-effect. In summary, based on the evidence presented, the difference in lower body power output between males and females, including masters athletes is inconsistent.

Activity Level

Impressively, the effect of activity level on lower-body power was found to be comparable to that of age based on zeroth order correlations presented in **Table 4.1**. For every unit increase in activity level (measured by the HUNT survey), the regression model predicted an additional 0.648W/kg of lower body power. As discussed previously above, males were predicted to gain 0.688W/kg for every unit increase in HUNT survey score, whereas females would gain 0.587W/kg. In 2021, Ramsey and colleagues carried out a 112-article systematic review investigating the effect of objectively measured PA and sedentary behaviour (SB) on strength and power in community dwelling adults aged 60 years or older. A majority of the 52 included studies used the sit to stand task as an indirect measure of lower body power output. While this cannot be directly compared to our data resulting from objectively measured lower body power output, the relationship between sit-to-stand power and PA is likely still meaningful. Each of the 52 studies investigated the effect of several different PA measures (steps, total PA, MVPA etc.) on lower body power (assessed via sit-to-stand). Of the total possible 90 outcomes across 52 studies, only two negative and not statistically significant effects of PA on muscular power were found. Thirteen of 52 studies did objectively measure maximal power, seven of those were assessed power via leg press and only Hartley (2018) and Edholm (2019) used jumping assessments. Edholm and colleagues (2019) measured squat jump performance and PA via Actigraph activity monitoring however the squat jump performance reported only specific force (N/kg) and not specific power (W/kg), limiting its utility for our comparison. Hartley and colleagues (2018) assessed power output via a Leonardo Mechanography GRF platform and PA via a triaxial accelerometer. Peak power was positively related to low and medium impacts (measured in g's), but curiously, not to high impacts. The lack of relation between high-impact activity and power may be explained by the "rarity of counts" recorded in that category (Hartley et al., 2018). As well, high effort activities are typically sustained for lower durations therefore it is reasonable to expect fewer counts. An important note related to our cohort is that active participants were 12-years younger on average and therefore the effect of activity level on power output may be confounded by the effect of age. In summary, evidence suggests a clear positive effect of PA on lower body power output, with the strength of that relationship dependent on the measurement methods for PA and lower body power.

Joint Power

Our joint power regression model showed that joint powers do decrease with age, and that activity level, as measured by the HUNT survey, contributes significantly to the model at the knee and hip only. Our data suggest that activity level failed to contribute significantly to prediction of ankle power. This may be the result of ankle power not necessarily being maximized during the CMJ, or the result of the knee and hip extensors receiving more attention in strength training efforts and as a result, power at these joints deteriorates at a lesser rate. Our cohort was comprised of several masters athletes, who on average, showed greater PA level (see annual percentage section). This group has likely been trained to utilize the large hip and knee extensor muscles in running and jumping events. This training may explain the activity level effect on knee and hip power.

Contrary to joint power during the maximal effort CMJ, joint powers during the submaximal functional tasks were largely unsuccessfully predicted by age, sex or activity level, with exception to hip power during the chair rise task. These results can be explained in part by our cohort's physical condition. All of our participants were able to complete our functional tasks whereas other literature has previously reported cases of older adult participants being unable to complete a chair rise task (Shea et al., 2018) or a stair climb task (Martha et al., 2017). This suggests that our cohort was more capable and mobile than some populations of older adults. It is possible that differences in joint power during submaximal functional tasks would be observed in less capable older adults (i.e. dependent living) who are operating closer to their maximal capacity. As is discussed in the limitations section below, the requirement for physical exertion in this study likely limited participation to those who were more capable than average. Not to mention, we included a large number of masters athletes in order to observe the effect of exercise quantity and type on power output. These athletes typically perform well above average into their 10th decade of life in some exceptional cases.

Power Normalized to Muscle CSA

The inclusion of lower body motion capture allowed us to calculate joint power during the CMJ, and by measuring muscular cross-sectional area (CSA) via MRI, we were able to evaluate knee power per unit CSA of knee extensor muscle. When knee power was normalized to knee extensor CSA, the effect of age on power output deteriorated. As discussed earlier, power normalized to body mass decreases linearly with age, so this finding was noteworthy. Additional inquiry determined that bodyfat levels were significantly greater in underactive adults over the age of 50 years. This may help explain why power normalized to body mass decreases with age – it is apparent that older adults, especially those who are underactive, retain more bodyfat, which is counterproductive to CMJ performance. Of note, our muscle CSA measurements are calculated with intramuscular fat removed and thus measures only lean mass. Therefore, the effects of additional passive tissue in older, underactive participants are largely controlled for. In 2004, Runge and colleagues measured power output via vertical jump (with arm swing, as mentioned above) in addition to calf muscle CSA (lean mass only) via computer tomography (CT). Contrary to our null findings, they found a significant correlation between age and power normalized to calf muscle CSA of -0.70 ($p<0.001$) in women and -0.81 ($p<0.001$) in men. They also found that the relationship between power normalized to body mass was stronger, with a correlation of -0.81 ($p<0.001$) in women and -0.86 ($p<0.001$) in men. This analysis differs from ours in a few important ways. First, they measured whole calf muscle CSA, which includes the ankle flexor musculature which does not drive CMJ performance to the same extent of the extensor musculature. Secondly, they presented whole-body power whereas we have assessed knee power relative to knee extensor CSA. It seems most reasonable to normalize power at the knee to the muscle CSA effectuating that joint. In 2000, Martin and colleagues measured power output during a maximal 3-4 second cycle ergometer in participants aged eight to 70 years. A subset of 24 subjects underwent MRI imaging of the thigh to validate a simpler calculation for lean thigh volume (LTV) based on skinfold and various thigh diameter measurements. These two figures allowed the authors to present power normalized to LTV. Power normalized to body mass showed a decrease of 30% between young adults (aged 20-29) and older adults (aged 60-70) whereas power normalized to LTV showed a decrease of only 19% over the same time. To summarize, it appears that when power is normalized to lean knee extensor CSA, lean calf muscle CSA, or LTV, the common finding was a reduced age effect. In fact, in the case of our data, no significant age effect was found at all.

The greatest limitation in this study is that of the cross-sectional design where we analyzed power in different people at different time points in their lives. A cohort study design would have observed lower body power in a single group of people, potentially separated into treatment and control groups (active versus underactive), as they aged. The main benefit of the cohort design is that the same person is being compared at 30 and 80 years of age. In our study design, the 30 and 80-year-old participants were different people with different genetics, diets and PA histories, to name a few confounding factors. Although the cohort design is more powerful than the cross-sectional design, time constraints led us to use the cross-sectional design. Another similar but largely unavoidable issue is that of the selection (or sample) bias. This bias suggests that the older adults in our sample population, including the non-athletes, are likely healthier and more physically capable than the age-matched population average (Metter et al., 1997). This effect is probably present since the less-healthy individuals in the population would not likely volunteer to participate in a study that requires substantial physical exertion.

Some inferences from the regression analyses can be complicated to interpret. For example, the linear regression analysis shown in **Figure 4.1** suggests that the rate of power loss is constant throughout an individual's lifespan. The rate of power loss probably changes throughout the lifespan in response to deteriorating physiology as discussed in the literature review (insufficient mechanical growth signaling, anabolic resistance etc.). However, the linear relationship is commonly depicted in the literature. This may be an artefact of the cross-sectional design, and the sampling bias mentioned above. Another complicated interpretation of the regression analysis involves the exemplar 30 and 80-year-old comparison presented above in **Figure 4.2**. The regression analysis suggests that an active 80-year-old can have approximately the same lower body power as an inactive 30-year-old. The caveat however is that this 80-year-old cannot expect to have such a high power at age 80 if they begin exercise training at age 80. Most individuals at this age will not be capable of such a high level of strenuous activity unless they have grown accustomed to it through training over many years. Regarding the question of early versus late-starting masters athletes, at least one study found no difference in vertical jump specific power between early starting (50 adulthood years of training) and late starting (19 adulthood years of training) masters athletes with an approximate mean age of 70 years (Piasecki et al., 2019). However, 19 years of training is still likely more than most which may explain the null finding. Still, it is probably reasonable to think of one's power output as a savings account with compounding interest, where starting early will have disproportionate effects later in life. This regression analysis does not clearly represent the effect of gaps in training, or the effect of timing of PA on one's power output.

Revisiting Hypotheses

1.1 Lower-body power during the CMJ would decrease with age but increase with activity level.

Accepted.

1.2 Joint power during the CMJ and functional tasks would decrease with age but increase with activity level.

Partially accepted. In the CMJ, knee and hip power decreased with age and increased with activity level. In the chair rise, hip power increased with activity level only.

Conclusion to RQ #1

Age, sex and PA measured by the HUNT questionnaire were predictive of lower-body power in the CMJ. Remarkably, the magnitude of effect for activity level approached that of age itself. Activity level had a positive protective effect on lower-body power in addition to knee and hip joint power during the CMJ however ankle power during the CMJ was not successfully predicted by activity level. This may be due to the large number of masters athletes in our cohort who presumably do more resistance training on these larger muscle groups. In general, the functional tasks were not well predicted by age, sex or PA level, implying that older adults of varying activity level do not use more or less specific joint power to complete daily tasks. The power “loss” through the lifespan was comparable to existing literature (approx. 50%) but the annual “loss” rates we found to be dependent on the reference age, type of jump and the technology used to measure power. Overall, PA measured by the HUNT survey was found to be remarkably protective of lower-body power in older adults and as a technical consideration, reference age is critical when discussing power loss in older adults.

5 Research Question #2: What is the effect of age, sex and athletics discipline on lower-body specific power and specific joint power during all tasks?

Introduction

Research Question #1 presented data to show that losses in lower-body power occur with age, but that these losses can be mitigated by PA, as measured by the HUNT survey. Many of our participants were also masters athletes from various athletic disciplines, focusing on very particular forms of exercise and training. This made for a convenient natural cohort in which we were able to assess the type of exercise that is most protective of muscular power in older adults. Of principal interest was the comparison between non-athletes, long-distance athletes and short-distance athletes. We predicted that short-distance masters athletes would have the greatest protection from losses in muscular power on the basis that their high acceleration and high-speed training and competition would protect them to the greatest extent. For the purpose of this analysis, short distance events included sprinting (50-400m), hurdles, pentathlon and decathlon and long-distance events were race-walking or running more than 3km.

Hypotheses

2.1 Lower-body specific power during the CMJ would decrease with age and would be highest in short distance athletes compared to all others.

2.2 Joint power during the CMJ would decrease with age and would be highest in short distance athletes compared to all others.

Methods

The sample size for Hypothesis 2.1 was 62 participants including 11 long-distance athletes (mean age 59.0), ten short-distance athletes (mean age 65.3), 32 non-athletes (mean age 55.3) and nine other athletes (mean age 43.4). The lower mean age of the “other” athletes is taken into consideration in the results and interpretation. The average sample size for Hypothesis 2.2 was 58, with stair descent hip power having the lowest sample size of 48 due to motion capture volume complications. The protocol for Hypothesis 2.1 and 2.2 consisted of CMJs, chair rises, stair ascents and stair descents performed with motion capture and force plates facilitating the calculation of lower body power via net force and pelvis CoM velocity, and inverse dynamics for joint power calculation. Athletics discipline was determined by asking participants whether or not they fit the definition of a masters athlete, or athlete, which were both listed on the intake form (See section 3.1 for the definitions or **Appendix D** for the intake form itself). Hypotheses 2.1 and 2.2 were assessed with multiple linear regression. More detailed methods for this research question can be found in common methods section.

Results

Multiple linear regression was used to determine the extent to which age, sex and athletics discipline predicted specific lower-body power (Hypothesis 2.1). Collectively, these variables generated a statistically significant predictive model of lower-body specific power ($F_{3,53}$ = 30.852, $p<0.001$, R^2=0.654). Age and athletics discipline made statistically significant contributions to the model ($p<0.001$), whereas sex did not (p=0.075). This relationship is shown below in **Figure 5.1** and additional model parameters are provided in **Table 5.1**. Note as well that when regression models were created to represent each athletic discipline (the three series of colored lines), the “other” athlete group model was not statistically significant, and a significant sex effect was found in the non-athlete group only. There was only one female short distance athlete, so no sex effects were represented in **Figure 5.1**.

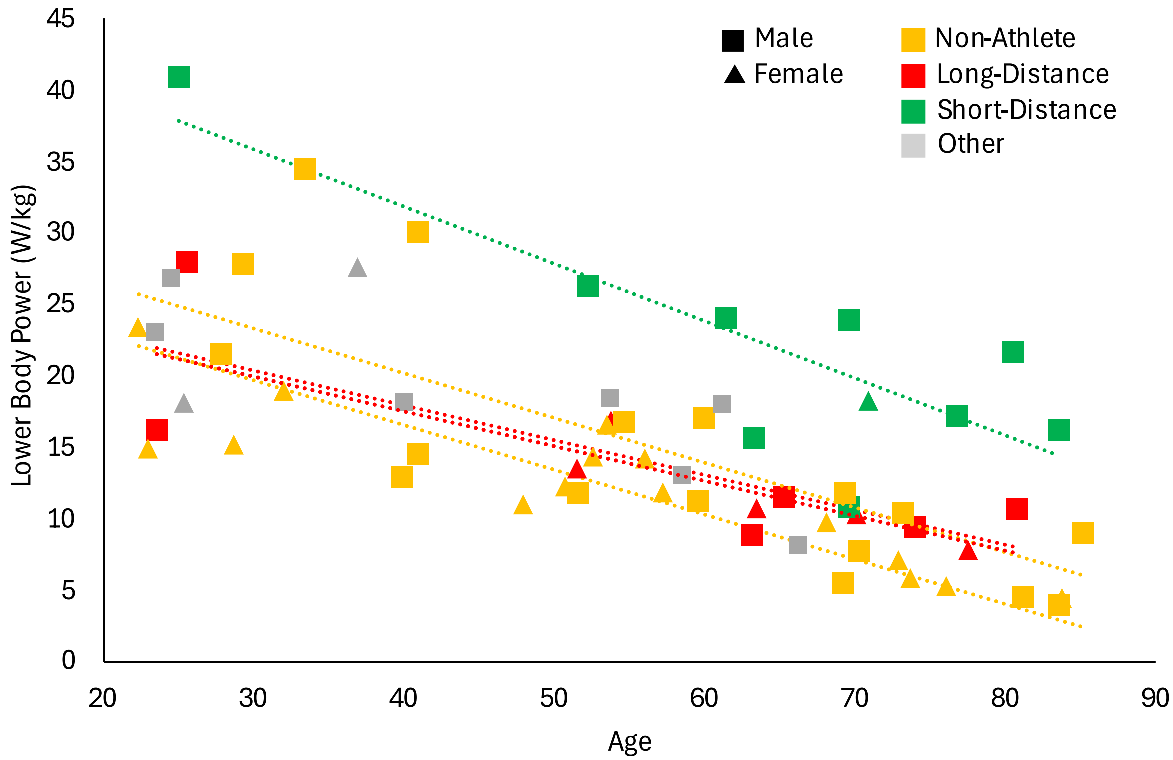

Figure 5.1 – The effect of age, sex, and athletics discipline on lower-body power output. Age and athletics discipline generated a statistically significant predictive model of specific muscular power output ($F_{3,53}$ = 30.852, p<0.001, R^2=0.654). Short-distance masters athletes show a large surplus in power compared to non-athletes and long-distance athletes. A male 80-year-old short-distance athlete is predicted to have the same approximate power output as a 50-year-old long-distance athlete or non-athlete. Athletics discipline was determined via survey. Two trendlines per athletic discipline represent the regression lines for male (top) and female (bottom). The greater sex effect is evident in non-athletes with a larger gap between trendlines, which may represent an interaction effect in the regression model (not tested). No trendline for female short-distance athletes due to insufficient data and no trendline for other athletes due to unsuccessful linear model.

Table 5.1 – The linear regression models for the various power assessments during the CMJ. Age, sex and athletics discipline generated a statistically significant model for lower-body specific power during the countermovement jump with all independent variables making statistically significant contributions to the model. For joint-level regression analyses, only age made a consistent, statistically significant contribution to the model. Correlations presented are zero-order.

CMJ					
DVs	**IVs: age, sex and athletics discipline (non=0, long=1, other=2, short=3)**				
Lower-body power, non-athletes	**Model**	$F_{2,31}$ = 28.585, **p<0.001**, Intercept = 32.719, R^2=0.640			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.313	0.042	**<0.001**	-0.776
	Sex	-3.668	1.600	**0.029**	-0.169
Lower-body power, long-distance	**Model**	$F_{2,10}$ = 8.559, **p=0.010**, Intercept = 27.710, R^2=0.602			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.244	0.061	**0.004**	-0.825
	Sex	-0.426	2.244	0.854	-0.215
Lower-body power, other	**Model**	$F_{2,8}$ =3.760, p=0.087, Intercept = 30.289, R^2=0.408			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.264	0.109	0.053	-0.744
	Sex	0.884	4.206	0.841	0.356
Lower-body power, short distance	**Model**	$F_{2,9}$ = 7.420, **p=0.019**, Intercept = 47.97, R^2=0.588			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.402	0.106	**0.007**	-0.823
	Sex	-1.207	5.663	0.837	-0.141

An additional regression model was created to determine if age, sex and athletic discipline were predictive of specific joint power during the CMJ (**Table 5.2**; Hypothesis 2.2). Collectively, these variables generated three statistically significant predictive models of specific ankle power ($F_{3,50}$ = 23.874, p<0.001, R^2=0.604), knee power ($F_{3,50}$ = 16.748, p<0.001, R^2=0.517) and hip power ($F_{3,50}$ = 11.863, p<0.001, R^2 = 0.431) during the CMJ. Age and athletics discipline contributed significantly to each model, whereas sex only contributed significantly to the ankle power model.

Table 5.2 – The linear regression models for joint-level power assessments during the CMJ. For joint-level regression analyses, only age made consistent, statistically significant contribution to the model. Correlations presented are zero-order.

CMJ					
DVs	**IVs: age, sex and athletics discipline (non=0, long=1, short=2)**				
Ankle power	**Model**	$F_{3,50}$ = 23.874, **p<0.001**, R^2=0.604, Intercept = 13.276			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.099	0.012	**<0.001**	-0.681
	Sex	-1.089	0.474	**0.026**	-0.213
	Athletics Discipline	0.820	0.295	**0.008**	0.149
Knee power	**Model**	$F_{3,50}$ = 16.748, **p<0.001**, R^2=0.517, Intercept = 11.411			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.105	0.016	**<0.001**	-0.564
	Sex	-0.889	0.632	0.166	-0.182
	Athletics Discipline	1.477	0.393	**<0.001**	0.285
Hip power	**Model**	$F_{3,50}$ = 11.863**, p<0.001**, R^2 = 0.431, Intercept = 8.325			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.072	0.014	**<0.001**	-0.488
	Sex	-0.508	0.534	0.346	-0.153
	Athletics Discipline	1.188	0.332	**<0.001**	0.307

Three additional regression models were used to determine if age, sex and athletics discipline were predictive of specific joint power during the chair rise, stair ascent and descent task. The regression model for knee power during chair rise was statistically significant however only athletics discipline made a significant contribution to the model, and the variance explained was approximately 17% (**Table 5.3**). Similarly, the regression model for ankle power during the stair ascent was statistically significant however only age made a significant contribution to the model and the variance explained was also approximately 17% (**Table 5.4**). No statistically significant models could be generated for the stair descent joint-level data (**Table 5.5**).

Table 5.3 – The linear regression models for the joint power assessment during the Chair Rise. Correlations presented are zero-order.

Chair Rise					
DVs	**IVs: age, sex and athletics discipline (non-athlete=0, long=1, short=2)**				
Ankle power	**Model**	$F_{3,50}$ = 1.261, p=0.298, R^2 = 0.075			
Knee power	**Model**	$F_{3,50}$ = 3.122, **p=0.035**, R^2 = 0.166			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	-0.005	0.016	0.082	-0.158
	Sex	-0.125	0.632	0.288	-0.201
	Athletics Discipline	0.167	0.393	**0.025**	0.303
Hip power	**Model**	$F_{3,50}$ = 1.172, p=0.331, R^2 = 0.070			

Table 5.4 – The linear regression models for the joint power assessment during the Stair Ascent. Correlations presented are zero-order.

Stair Ascent					
DVs	**IVs: age, sex and athletics discipline (non-athlete=0, long=1, short=2)**				
Ankle power	**Model**	$F_{3,47}$ = 3.062, **p=0.038**, R^2 = 0.173			
	Coefficients	**β**	**SE**	**p**	**r**
	Age	0.009	0.003	**0.006**	0.368
	Sex	0.101	0.114	0.378	0.117
	Athletics Discipline	-0.057	0.071	0.421	-0.064
Knee power	**Model**	$F_{3,47}$ = 0.927, p=0.436, R^2 = 0.059			
Hip power	**Model**	$F_{3,47}$ = 1.033, p=0.387, R^2 = 0.066			

Table 5.5 – The linear regression models for the joint power assessment during the Stair Descent. Correlations presented are zero-order.

Stair Descent		
DVs	**IVs: age, sex and athletics discipline (non-athlete=0, long=1, short=2)**	
Ankle power	**Model**	$F_{3,48}$ = 1.593, p=0.204, R^2 = 0.096
Knee power	**Model**	$F_{3,47}$ = 0.493, p=0.689, R^2 = 0.033
Hip power	**Model**	$F_{3,42}$ = 0.988, p=0.408, R^2 = 0.071

Additional Findings

As with Research Question #1, a categorical analysis was conducted as well. A three-way ANOVA was used to assess the effect of sex (male or female), age (<30, 30-50, 50-70, >70) and athletic discipline (non-athlete, long, short or other) on lower-body specific power output. The analysis revealed no interaction effects (p>0.123, η_p^2 < 0.133) but did reveal main effects for age ($F_{3,61}$ = 14.034, p<0.001, η_p^2 = 0.513; **Figure 5.2**) and athletics discipline ($F_{3,61}$ = 7.594, p<0.001, η_p^2 = 0.363; **Figure 5.3**). Post-hoc analysis on age groups with Bonferroni correction revealed that adults aged 30 or younger produced significantly more power than 50-70 (Δ = 10.808W/kg, p<0.001) and >70-year-olds (Δ = 13.364W/kg, p<0.001), and that 30–50-year-olds produce significantly more power than the 50-70-year-olds (Δ = 6.949W/kg, p=0.004) and >70-year-olds (9.505W/kg, p<0.001).

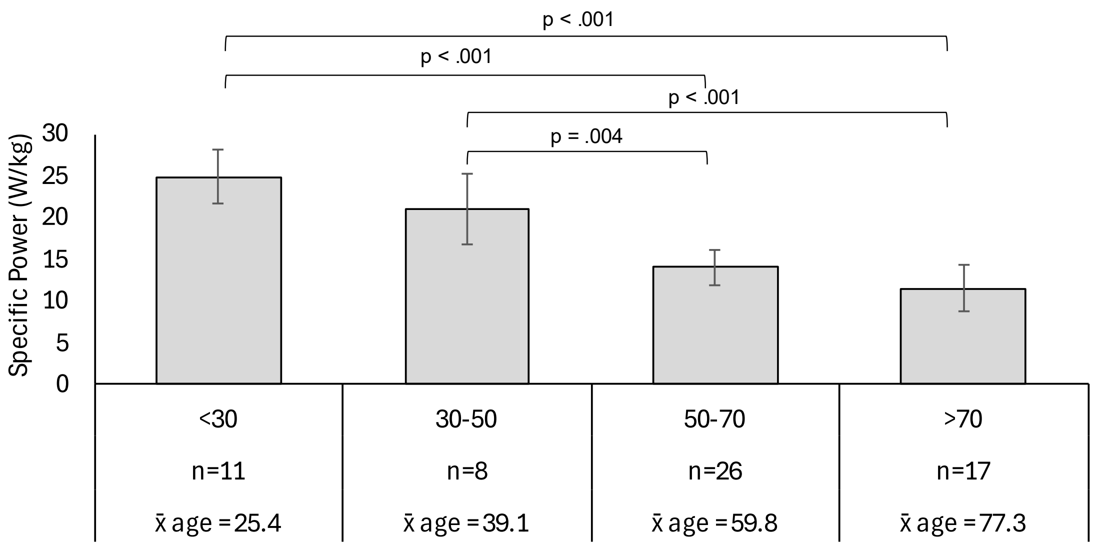

Figure 5.2 – The relationship between age and specific lower-body power output. A three-way ANOVA determined that adults aged less than 50 years of age produced greater power than those aged greater than 50 years of age. Error bars represent 95% confidence intervals.

Post-hoc analysis with Bonferroni correction on athletics discipline revealed that short distance athletes produced significantly greater power during the CMJ compared to both non-athletes (Δ = 9.582W/kg, p<0.001) and long-distance athletes (Δ = 11.419W/kg, p<0.001; **Figure 5.3**). Additionally, 'Other' athletes had significantly greater power than long distance athletes (Δ = 7.565W/kg, p=0.021). Athletes in the 'Other' category competed in CrossFit (3), middle-distance running (2), field hockey (2), rugby (1) and Dragonboat (1). An important caveat though is that the average age in the Other athlete group was lower than the average age in all other groups by at least 12 years (compared to non-athletes) and at most 22 years (compared to short-distance athletes). Considering the age-effects shown above (i.e., a loss of 0.279W/kg per year of age), the greater power between 'other' athletes and non-athletes should be interpreted cautiously.

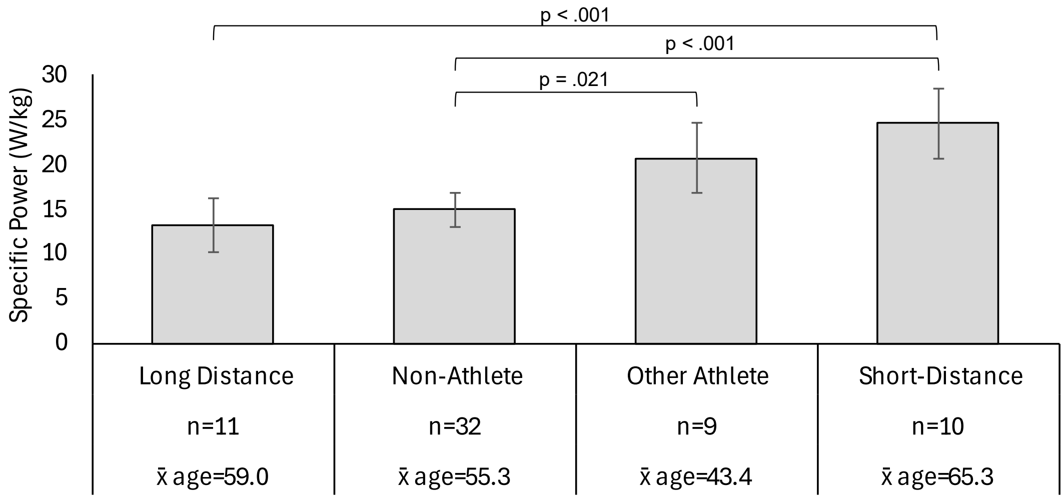

Figure 5.3 – The relationship between athletic discipline and specific lower-body power output. A three-way ANOVA showed that short distance athletes produced significantly more power than non-athletes and long-distance athletes. As well, 'Other' athletes produced significantly greater power than non-athletes. Other athletes include those from CrossFit (3), middle distance running (2), field hockey (2), rugby (1) and Dragonboat (1). Error bars represent 95% confidence intervals.

Linear regression assessed the effect of athletic history on muscular power output. Recreational physical activity hours, competitive physical activity hours or a combination thereof (determined via survey) had no significant effect on muscular power output ($F_{2,54}$ = 2.062, p=0.137). A statistical comparison between early starting (defined as a current masters athlete who began competition *prior* to 50 years of age; Piasecki et al., 2019) versus late starting master athletes (defined as a current masters athlete who began competition *after* 50 years of age) was not feasible due to low sample size (20 early starting and 4 late starting athletes). Nonetheless, a cursory analysis of means showed average power output of 15.9W/kg for early starting athletes (n=20, range: 5.7 to 41W/kg) and 14.9 for late starting masters athletes (n=4, range: 10.3 to 17.3W/kg). A caveat, however, is that late starting athletes were also 73.4 years of age on average versus 62.6 for early starting masters athletes.

Four separate three-way MANOVAs were used to analyze specific joint power in each task (CMJ, chair, ascent, descent). Each MANOVA assessed the effect of age (<30, 30-50, 50-70, 70+), sex (male or female) and athletics discipline on peak specific joint power (ankle, knee, and hip).

MANOVA testing for the CMJ revealed a significant interaction between age and athletics discipline ($F_{3,58}$ = 3.169, η_p^2 = 0.334, p=0.013). Post hoc testing with Bonferroni correction revealed that knee power was greater in non-athletes aged less than 30 years versus those aged 50-70 years (Δ = 4.015W/kg, p=0.006), and those aged greater than 70 years (Δ = 5.465W/kg, p<0.001). Post hoc testing also revealed that knee power was greater in short-distance athletes aged less than 30 years versus those aged 50-70 years (Δ = 11.895W/kg, p<0.001) and greater than 70 years (Δ = 12.649W/kg, p<0.001).

A three-way MANOVA investigating joint power during the chair rise task detected a statistically significant interaction effect on age and athletics discipline on hip power ($F_{4,51}$ = 2.599, p=0.033, η_p^2 = 0.291). Follow up post hoc analysis failed to identify specific group differences, likely resulting from our cohort lacking 30–50-year-old long distance and short distance athletes. An additional three-way MANOVA investigating joint power during the stair ascent task failed to detect a significant interaction (p>0.271, η_p^2 < 0.101) or main effect of age, sex or athletics discipline on power at any of the three lower-limb joints (p>0.187, η_p^2 < 0.055). A final three-way MANOVA investigating joint power during the stair descent task detected a significant interaction effect between age and athletics discipline on hip power ($F_{5,47}$ = 3.718, p=0.010, η_p^2 = 0.399). Follow up post hoc testing with Bonferroni correction revealed that long distance athletes less than 30 years of age produced less hip power compared to those aged 50-70 (Δ = 5.920W/kg, p=0.002) and those aged greater than 70 years (Δ = 4.225W/kg, p=0.039).

Key Findings

- Cross-sectional data in this cohort predicted that non-athletes would lose 0.313W/kg per year of lower body power per year, compared to 0.244W/kg per year for long-distance athletes and 0.402W/kg per year for short-distance athletes.
- Short distance athletes produce significantly more power than non-athletes and long-distance athletes, despite older average age in that cohort.

- In the CMJ, non-athletes and short-distance athletes aged <30 produce greater knee power than those aged >50 years.
- Long distance athletes aged >50 produce more hip power during stair descent compared to those aged <30 years.

Discussion

Overview

This section discusses the results of Research Question #2 in the context of existing literature related to rates of power loss in short and long-distance masters athletes. Short-distance athletes showed a greater y-intercept and steeper slope which implies higher power output throughout the lifespan, despite degrading at a greater rate by absolute measure.

Absolute Rates by Athletic Discipline

Linear regression analysis found that specific power decreased at a rate of 0.279W/kg per year in our whole cohort. Separate models for each athletic discipline found that specific power decreased at a rate of 0.313W/kg per year in non-athletes, 0.244W/kg per year in long-distance athletes and 0.402W/kg per year in short-distance athletes. Our dataset also included nine “other” athletes consisting of CrossFit and Rugby athletes, among others but this group failed to form a statistically significant regression model. A 2021 study by Alvero-Cruz and colleagues measured power output in 135 power athletes, 40 mixed athletes and 69 endurance athletes. Their regression model predicted reductions in power output of 0.485W/kg per year in power athletes and 0.200W/kg per year in endurance athletes. These regression slopes are approximately 20% greater than shown in our models. This difference may be related to the free use of arms allowed by Alvero-Cruz and colleagues. Although conceivably, the additional power gained by using arm swing during the CMJ would affect the entire cohort and thus should not affect the slope, unless arm swing does not benefit older adults to the same extent as younger adults. This is possible as arm swing during CMJ is suspected to alter the force-velocity characteristics of the lower limb muscles (Lees et al., 2004) – a change which may not be well tolerated by older adults. In summary, based on our finding and the work of Alvero-Cruz and colleagues (2021), approximate absolute per annum losses in power of 0.20 to 0.25W/kg per year is predicted in long-distance athletes, and 0.40 to 0.50W/kg per year is predicted in short-distance athletes. As will be shown below, despite the steeper slope, the per annum percentage losses between long- and short-distance athletes are still equitable.

Percentage Rates by Athletic Discipline

In 1991, Grassi and colleagues conducted an investigation that was remarkably similar to our own. They evaluated lower limb power output via vertical jump (hands on hips) using force plates in both masters endurance and power athletes plus controls, all aged 40-78 years. They also measured body fat (skin fold whereas we used BIA) and approximated lean thigh volume via anthropometric measures (whereas we used MRI). Perhaps the greatest distinction between their work and ours was our inclusion of activity level measurement and their use of a squat jump (no countermovement). Despite these differences, their findings were very similar to ours. Firstly, they found an approximate 50% decrease in power between 20- and 75-year-old masters athletes (0.91% per year), regardless of discipline. Between the same ages in long distance athletes, our regression model predicted losses of 58.8% (1.07% per year) in males and 59.9% (1.09%) in females. In short distance athletes, our model predicted losses of 55.4% (1.01% per year) in males and 57.1% (1.04% per year) in females. Grassi and colleagues (1991) did not present the losses for their control participants, however our data suggests inflated losses in this group - 65% (1.34% per year) in males and 79.6% (1.59% per year) in females. In 2002, Pearson and colleagues evaluated power output via a Nottingham Power Rig in 54 elite masters athlete weightlifters aged 40 to 87, with 54 untrained control participants. They found a peak power loss of 1.3% per year relative to that of a 45-year-old, compared to 1.2% per year in controls. This finding is contrast with our data which show greater per annum losses in our non-athlete cohort compared to the losses in our short-distance cohort, and long-distance cohort. Short-distance athletes were predicted to lose 1.12W/kg per year in males and 1.16W/kg per year in females respectively, versus 1.20% and 1.22% per year in male and female long-distance athletes, and 1.34% per year and 1.59% per year in male and female non-athletes. The discrepancy between our results and those of Pearson and colleagues may be explained in part by the lack of participants over the age of 70 in the control group of Pearson and colleagues (2002). As well, their per annum data are calculated with a 45-year-old reference, which will inflate the per annum loss rate as discussed above in RQ#1. In summary, our data, and that of both Grassi (1991) and Pearson (2002) suggest approximate power loss of 50-60% over the course of 50 years in long- and short-distance masters athletes. Our data also suggests that non-athletes are predicted to lose a greater proportion of their 30-year-old reference power (67.1% for males and 79.6% for females) by age 80. **Table 5.6** provides a summary of some existing literature relating to power loss in masters athlete and control populations.

Table 5.6 – A summary of masters athlete lower-body power literature segregated by different athletics disciplines.

Reference	Sample size	Age	Absolute Loss per Year	Percentage Loss (%)	Percentage Loss per Year
Alvero-Cruz et al., 2021 ‡					
Power	135	56 (median)	0.485W/kg	39.3% @ 80 wrt 40*	0.98%*
Endurance	69	57 (median)	0.200W/kg	23.1% @ 80 wrt 40*	0.58%*
Grassi et al., 1991 §					
Power	48	17.7-74.3*	Undisclosed	~50% @ 75 wrt 20	0.92%*
Endurance	87	19.0-74.3*	Undisclosed	~50% @ 75 wrt 20	0.90%*
Control	36	23.0-64.9*	Undisclosed	43% @ 65 wrt 25*	1.08%*
Pearson et al., 2002 ¶					
Weightlifters	54 (54m)	45.4-84.3*	Undisclosed	49% @ 85 wrt 45	1.3%
Control	~54 (54m)	42.7-84.2*	Undisclosed	51% @ 85 wrt 45	1.2%
Michaelis et al., 2008 ‡					
Short-distance, male	135	56 (35-90)	0.59W/kg	39.7% @ 80 wrt 40*	0.99%*
Short-distance, female	120	55 (35-86)	0.50W/kg	40% @ 80 wrt 40*	1.00%*
Long-distance, male	120	60 (40-87)	0.52W/kg	45% @ 80 wrt 40*	1.13%*
Long-distance, female	81	55 (37-80)	0.27W/kg	29% @ 80 wrt 40*	0.73%*
This Research #					
Short-distance	10 (9m, 1f)	65.3 (25.1-83.6)	0.402W/kg	56.9% @ 80 wrt 30	1.14%†
Long-distance	11 (6m, 5f)	59.0 (23.6-80.8)	0.244W/kg	60.5% @ 80 wrt 30	1.21%†
Non-athletes	32 (17m, 15f)	55.3 (22.4-85.2)	0.313W/kg	73.3% @ 80 wrt 30	1.47%†
~Approximate *****Calculated from available data **†**Male and female average **‡**Leonardo, CMJ w/ arm swing **§**Force plate, squat jump w/o arms **¶**Nottingham power rig **#**Force plate, CMJ w/o arm swing					

Joint Power

Linear regression models investigating the effect of age, sex and athletic discipline on ankle, knee and hip joint power during the CMJ were statistically significant. Each model had significant contribution from age and athletic discipline, but only ankle power demonstrated a significant contribution from sex. All models predicted that short distance athletes would generate the highest power output at all three joints with the largest effect observed at the knee joint. These results suggest that short distance athletes have greater capacity for power at the joint level as well as lower-body power which was presented earlier. It is possible that such broad capacity across all joint avails these older adults to additional movement strategies which may confer functional benefit, although this remains to be tested.

Revisiting Hypotheses

2.1 Lower-body specific power during the CMJ would decrease with age and would be highest in short distance athletes compared to all others.

Accepted.

2.2 Joint power during the CMJ would decrease with age and would be highest in short distance athletes compared to all others.

Accepted.

Conclusion to RQ #2

Athletic discipline had a positive protective effect on lower-body power in addition to ankle, knee and hip joint power during the CMJ with short-distance athletes having the highest predicted power. In contrast, short-distance masters athletes also had the steepest rate of power “loss” compared to long distance and non-athletes. The implication is that short distance athletes are predicted to have higher power output at a given age but are also expected to lose power at a greater rate. This appears to be largely inconsequential though as their higher initial power output serves as a substantial buffer thereby sustaining higher power output well beyond the normal lifespan. This finding agrees with existing literature showing approximate power “loss” of

50-60% over the course of 50 years in both long- and short-distance masters athletes, despite the steeper “loss” rate also observed in this population. Overall, short-distance athletes were shown to produce the greatest lower-body power at all ages which suggests that this form of athletics may serve as an enjoyable and social form of PA with the additional benefit of preserving power and presumably, function. Future research should determine if the greater joint power expressed by short distance athletes during the CMJ, facilitates the performance of functional tasks requiring a fraction of this power.

6 Research Question #3: What is the combined effect of age, sex and activity level on changes in lower-body power output?

Introduction

Fatigue can be defined as a reduction in the ability of a muscle to produce force or power, whether or not the task can be sustained (Enoka and Duchateau, 2008). Reductions in power are likely problematic for older adults due to the strong relationship between power and functional task performance in this cohort (Balachandran et al., 2022; Foldvari et al., 2000). Paradoxically, a majority of the literature indicates that older adults are equivalently or *less* fatigable in a variety of test conditions, i.e., isometric contractions (versus dynamic), sustained contractions (versus intermittent), using force or endurance as the index of fatigue, etc. However, an assessment using dynamic tasks with muscular power as the index of fatigue has been shown to be revealing of fatigue in older adults (Christie et al., 2011). This research question sought to quantify fatigue in a cohort of young and old adults, with a countermovement jump assessment repeated intermittently after bouts of functional tasks. Exercise status was quantified and included as a factor as it has been considered an often omitted confounding factor (Christie et al., 2011).

Hypothesis

3.1 Underactive older adults (>50 years) will show fatigue in the post condition, quantified by reduced power in the CMJ.

Methods

The sample size for Hypothesis 3.1 was 62 (35 active, mean age 50.8; 27 underactive, mean age 62.3). The average sample size for the kinematic analysis was 56 but in the case of trunk angle change during stair descent the sample size was 40 participants. The lower sample size was due to poor motion capture volume at the top of the stair apparatus when the participant was facing down the stair set. The protocol used to assess Hypothesis 3.1 involved the functional task circuit outlined in the common methodology section. The functional task consisted of chair rising, stair ascent, stair descent, and CMJs. Hypothesis 3.1 was addressed with a three-way ANOVA and subsequent additional supporting analyses were conducted with descriptive statistics, boxplots and linear regression. More detailed methods for this research question can be found in common methods section.

Results

Change in Power

A three-way ANOVA with age, sex and activity level as factors, found no significant interactions ($\eta_p^2<0.043$, $p>0.191$) or main effects ($\eta_p^2<0.108$, $p>0.174$; Hypothesis 3.1) on change in lower-body specific power. A 95% CI revealed that the average change in specific power was significantly greater than zero (0.291W/kg to 1.002W/kg; **Figure 3.1, left**) with a small effect size (Cohen's d=0.46). Likewise, a separate 95% CI revealed that jump height was also significantly different from zero (0.018 to 0.028m; **Figure 3.1, right**) with a large effect size (Cohen's d=1.11). These findings imply that on average, participants displayed a *net gain* in power and jump height between the first three and last three CMJ repetitions.

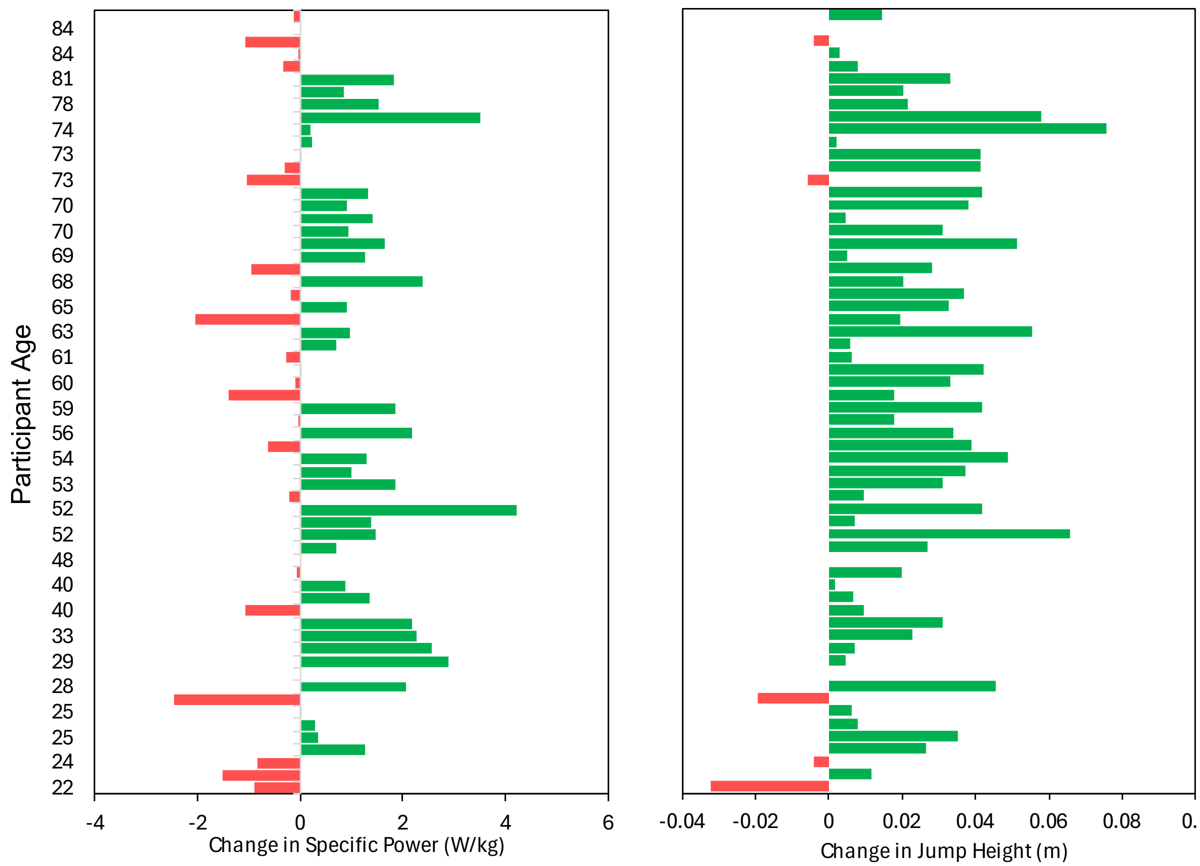

Figure 6.1 – The change in specific power (left) and jump height (right) from the first three to the last three CMJ. from the first three to the last three CMJ. A three-way ANOVA investigating the effect of sex, age and activity level on change in power failed to uncover any interaction or main effects (p>0.174).

Change in Kinematics

Although not driven by a specific hypothesis, changes in kinematics are likely important to document as they may have affected changes in power output directly by modifying jump technique, or kinematic changes and movement variability may have allowed for brief rest periods, thereby affecting the development of fatigue. The ‘pre’ period refers to the trials nearest to the beginning of the circuit (with the caveats mentioned in Methods – Circuit Event Processing) while the ‘post’ period refers to the trials nearest to the end of the circuit.

In the CMJ, at the instant of peak GRF, the greatest changes in joint angle from pre to post were observed at the left and right hip joints with RMSD values of 14.10° and 13.83°, respectively while the trunk showed the lowest RMSD value of 5.05° (**Table 6.1**).

Table 6.1 – Changes in joint angle in the CMJ at the instant of peak GRF.

	Ankle (L)	Ankle (R)	Knee (L)	Knee (R)	Hip (L)	Hip (R)	Trunk
RMSD (°)	4.98	5.43	11.32	11.94	14.10	13.83	5.05
Mean (°)	0.75 (ext)	0.59 (ext)	0.48 (flex)	0.08 (flex)	0.31 (ext)	0.40 (ext)	0.27 (flex)
SD (°)	5.06	5.49	11.61	12.04	14.22	13.94	5.08
Min (°)	-11.28	-14.15	-38.11	-31.65	-33.37	-29.99	-19.08
Max (°)	17.17	16.68	32.63	36.02	37.75	39.57	13.46
Range (°)	28.45	30.84	70.74	67.67	71.12	69.56	32.54

The small, positive mean joint angles shown above in **Table 6.1** indicate that at post, ankle joints tended towards greater extension, knee joints towards greater flexion, hip joints towards greater extension and the trunk towards greater flexion. No significant differences were observed via 95% CI at the $\alpha=0.05$ level of significance (see **Figure 6.10** for all task 95% CIs). These joint angle changes were not consistent as evidenced by the greater than 70° range exhibited at the left knee and hip (**Table 6.1**, above). Note that this range represents joint angle changes from two different participants – i.e., one participant produced 33° greater hip flexion at post (P002), and another produced 38° greater hip extension at post (P025), yielding a range of over 70°. The full distribution of all joint angle changes in the CMJ is shown in **Figure 6.2** with the greatest outlier participant codes indicated. The participants responsible for the 70° range mention above are indicated in the Hip (L) plot.

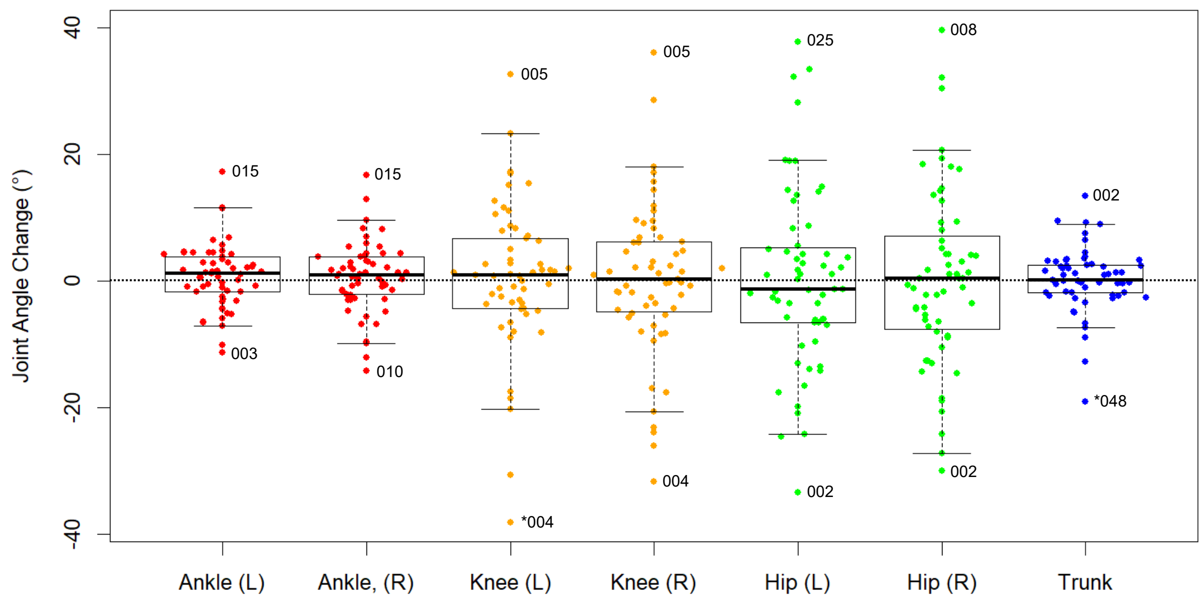

Figure 6.2 - Boxplot of joint angle change from the first three to the last three CMJ (pre to post). The variability between participants was large ranging upwards of 70 degrees at the knee and hip joints. 29 outliers (within 1.5 to 3.0 IQR), and two extreme outliers (identified by (*), beyond 3.0 IQR) were indicated. Data labels indicate the participants who displayed the greatest change.

For context, the precise pose of the participant (004) who showed the most change between pre and post timepoints across all joints, despite the same peak GRF event trigger is shown in **Figure 6.3**. This change was calculated as the mean of the absolute value of joint angle change per participant (and is therefore independent of direction). This participant showed 151° of joint angle change across seven joints – bilateral ankle, knee, hip, and trunk – while the average across the whole cohort was 48°.

Changes to joint angles during the CMJ were small on average, but large based on range. We investigated whether these joint angle changes may have affected initial power output, the change in power from pre to post, or the change in jump height from pre to post. Multiple linear regression analysis determined that these large and inconsistent joint angle changes during the CMJ (represented by RMSD in this part of the analysis) had no significant linear relationship with initial power output ($F_{1,58}$ = 0.709, p=0.403), to changes in power output from pre to post ($F_{1,58}$ = 1.893, p=0.174), or to changes in jump height ($F_{1,58}$ = 0.477, p=0.493). Change in CMJ technique did not affect initial power, change in power or change in jump height. As we will see next, joint angle changes in the functional tasks were lesser in magnitude but more consistent. Their connection to fatigue will also be investigated.

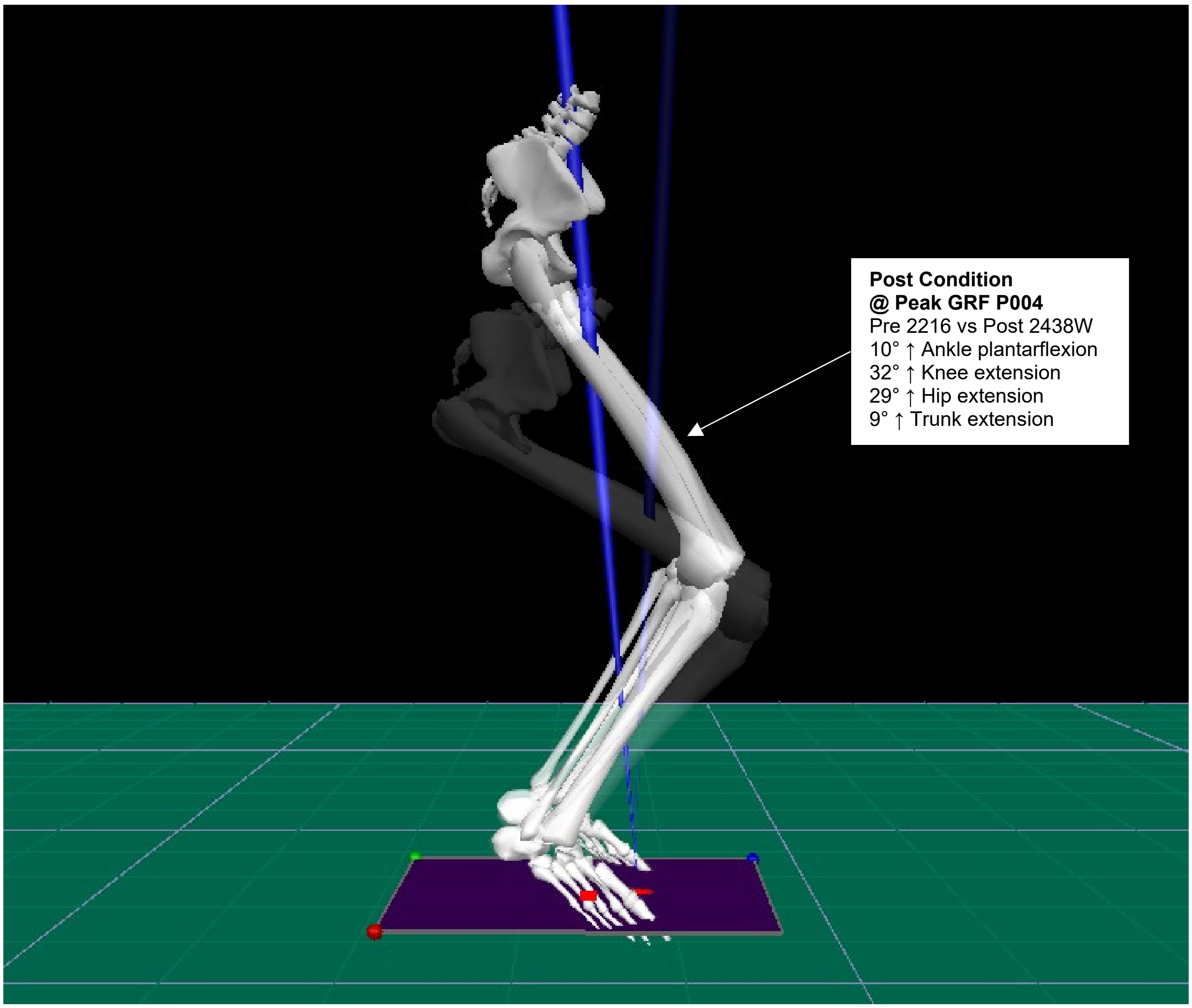

Figure 6.3 - 'Worst-case' pre and post postures during CMJ at peak GRF. The posture in the pre condition is shown as a shadow, while the post condition is shown in the forefront. This participant represented the largest change in joint angle (151°) across seven joints observed throughout the circuit. The joint angle data in the text box are averaged left and right.

In the chair rise task, at the instant of peak GRF, the greatest changes in joint angle from pre to post were observed at both the left and right hip joints with RMSD values of 5.52° and 5.05°, respectively while the trunk, once again, showed the lowest RMSD value of 2.86° (**Table 6.2**).

Table 6.2 – Changes in joint angle in the chair rise at the instant of peak GRF.* Indicates a significant difference from zero (α = 0.05). 'ext' or 'flex' indicate an extension or flexion dominance at post.

	Ankle (L)	**Ankle (R)**	**Knee (L)**	**Knee (R)**	**Hip (L)**	**Hip (R)**	**Trunk**
RMSD (°)	3.64	3.36	3.89	3.89	5.52	5.05	2.86
Mean (°)	0.64 (ext)	0.47 (ext)	-0.69 (ext)	-0.68 (ext)	0.76 (ext)	0.65 (ext)	**-0.79 (ext)***
SD (°)	3.68	3.39	4.01	3.89	5.66	5.09	2.77
Min (°)	-8.03	-11.01	-13.72	-12.44	-12.24	-10.24	-7.15
Max (°)	9.16	8.09	9.78	8.39	11.33	8.98	9.83
Range (°)	17.20	19.10	23.50	20.84	23.57	19.21	16.99

Based on mean joint angles, at post, the ankle, knee, hip and trunk all tended towards greater extension. Only the change in trunk angle was significantly different from zero with an average of 0.79° (95% CI: -0.09° to -1.50°) greater extension in the post condition (see **Figure 6.10** for all 95% CIs). The changes were more consistent than those observed in the CMJ, with a maximum range of approximately 23° observed at the left hip. The full distribution of all joint angle changes in the chair rise is shown in **Figure 6.4** with the greatest outlier participant codes indicated.

For context, a Visual3D model shows the precise pose of the participant (027) who showed the most change between pre and post timepoints across all joints, despite the same peak GRF event trigger (**Figure 6.5**). Similar to the CMJ, this change was calculated as the mean of the absolute value of joint angle change per participant (and is therefore independent of direction). This participant showed 51° of joint angle change across seven joints – bilateral ankle, knee, hip, and trunk – while the average across the whole cohort was 22°.

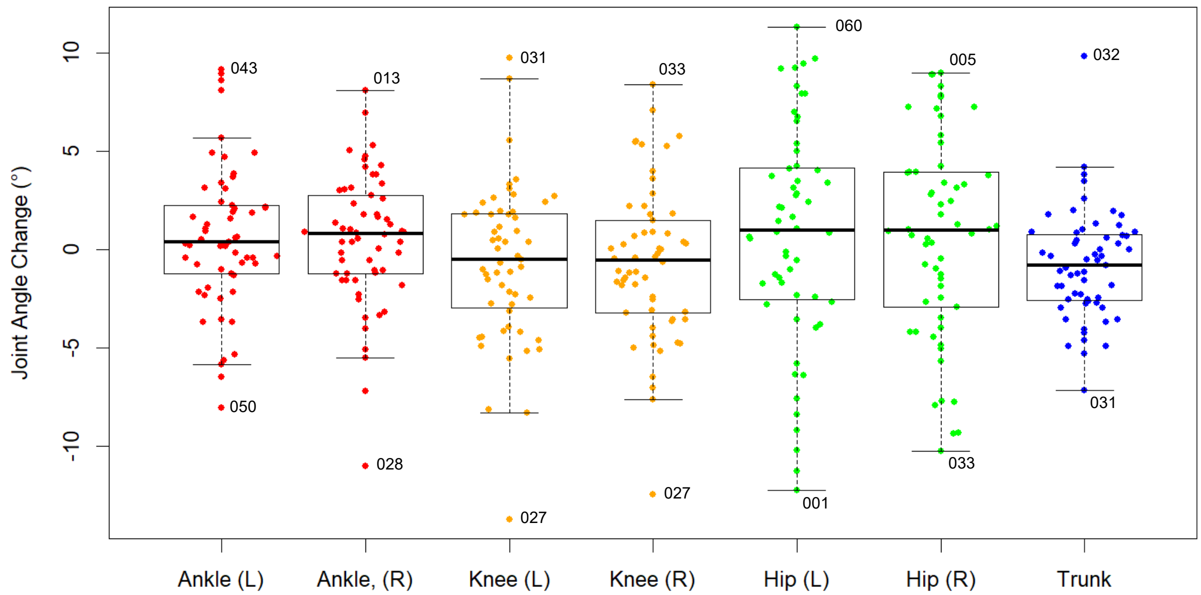

Figure 6.4 - Boxplot of joint angle change from the first three to the last three chair rises (pre to post). The changes were less variable than those observed in the CMJ with coefficients of variation ranging from 3.49° for the trunk to 7.85° for the right hip but were still consistent with broadly distributed data. As well, 12 outliers were indicated (within 1.5 to 3.0 IQR) with 8 of those occurring at the left and right ankles. 95% CI suggests that the change in trunk angle was significantly different from zero (-0.09, -1.50). Data labels indicate the participants who displayed the greatest change.

Joint angle changes during the chair rise were lesser in terms of range than the CMJ. Nonetheless, change in trunk angle between pre and post timepoints was statistically significant. Therefore, to determine if the joint angle change at the trunk had any effect on power or the change in power during the circuit, a linear regression analysis was conducted. Regression analysis determined that change in trunk angle during the chair rise had no significant linear relationship with initial power output in the CMJ ($F_{1,58} = 0.431$, $p=0.514$) or on change in power in the CMJ ($F_{1,57} = 0.303$, $p=0.584$). Therefore, the statistically significant change in trunk angle between pre and post had no effect on power output or change in power output in the CMJ.

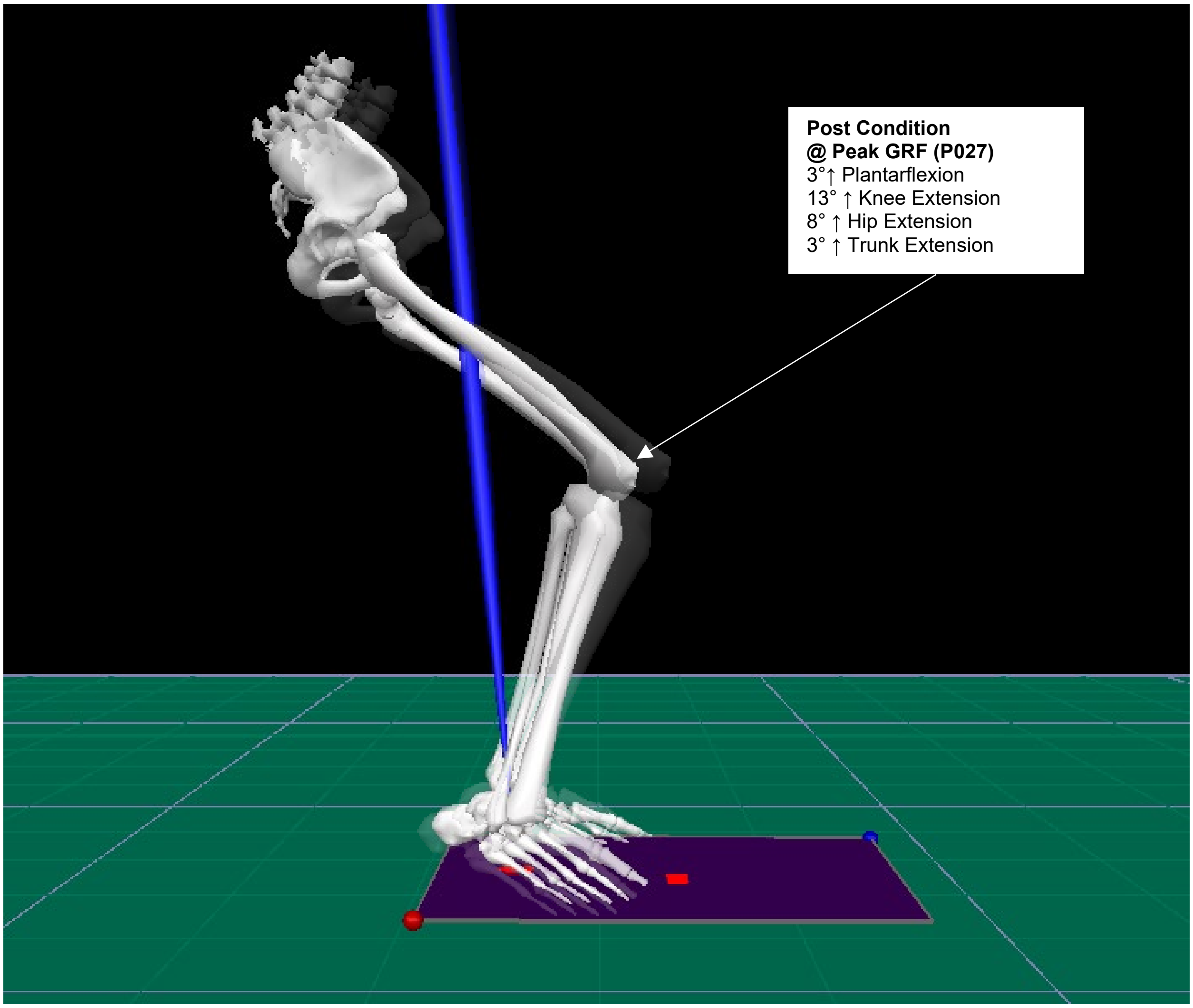

Figure 6.5 – 'Worst-case' pre and post postures during chair rise at peak GRF. The posture in the pre condition is shown as a shadow, while the post condition is shown in the forefront. This participant represented the largest change in joint angle (51°) across seven joints observed throughout the circuit. The joint angle data in the text box are averaged left and right.

In the stair ascent task, at the instant of initial contact, the greatest changes in joint angle from pre to post were observed at the knee with an RMSD value of 3.64° and the trunk showed the lowest RMSD value of 1.61° (**Table 6.3**).

Table 6.3 – Changes in joint angle in the stair ascent at the instant of peak GRF. * Indicates a significant difference from zero (α = 0.05). 'ext' or 'flex' indicate an extension or flexion dominance at post.

	Ankle	**Knee**	**Hip**	**Trunk**
RMSD (°)	2.85	3.43	3.64	1.61
Mean (°)	-0.59 (flex)	0.00 (---)	**-1.02 (flex)***	0.30 (flex)
SD (°)	2.87	3.49	3.56	1.61
Min (°)	-7.65	-8.47	-12.48	-4.02
Max (°)	6.99	7.38	9.05	3.69
Range (°)	14.64	15.85	21.54	7.70

Based on mean joint angles, between pre and post during the stair ascent, the ankle, knee, hip and trunk segment tended towards greater flexion. The change in hip angle was significantly different from zero with an average of -1.02° (95% CI: -0.11° to -1.93°) greater flexion in the post condition (see **Figure 6.10** for all task 95% CIs). The full distribution of all joint angle changes in the stair ascent is shown in **Figure 6.6** with the greatest outlier participant codes indicated.

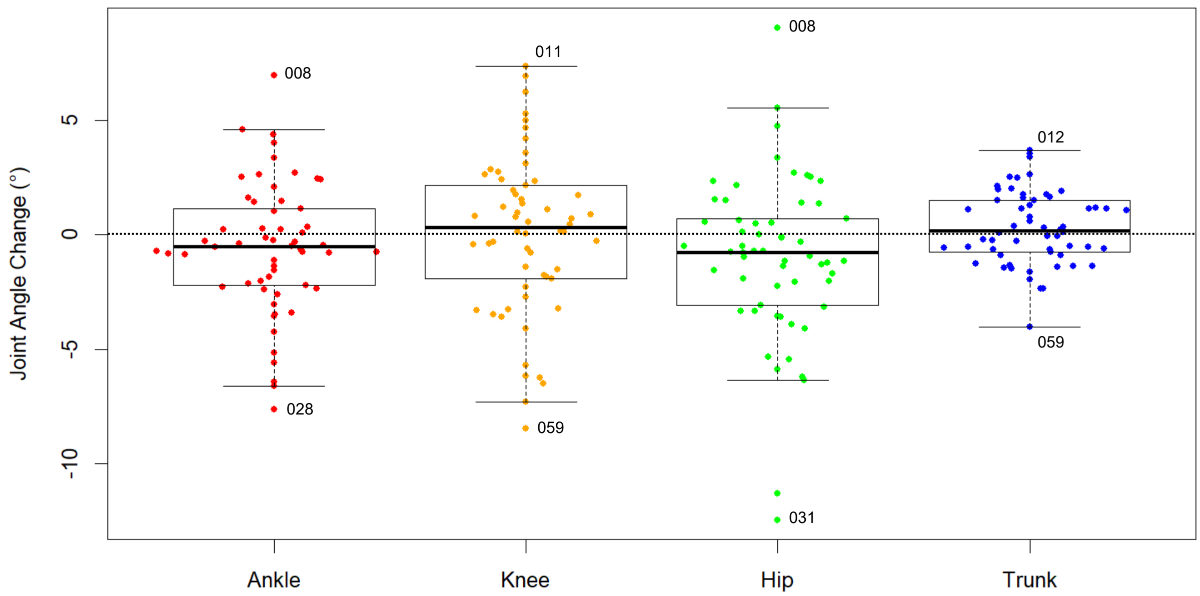

Figure 6.6 - Boxplot of joint angle change from the first two to the last two stair ascents (pre to post). On average, between pre and post during the stair ascent, the ankle, knee, hip and trunk joint tended towards greater flexion. However, the variability between participants was large with all joint-level coefficients of variation greater than 3.50 where 0.5 is typically considered a broad distribution. As well, 6 outliers were indicated (within 1.5 to 3.0 IQR), three of which occurred at the hip. 95% CI suggest that the change in hip angle was significantly different from zero (-0.11, -1.93). Data labels indicate the participants who displayed the greatest change.

For context, the precise pose of the participant (008) who showed the most change between pre and post timepoints across all joints, despite the same IC event trigger is shown in **Figure 6.7**. This change was calculated as the mean of the absolute value of joint angle change per participant (and is therefore independent of direction). This participant showed 24° of joint angle change across four joints –ankle, knee, hip, and trunk – while the average across the whole cohort was 9°.

Joint angle changes during stair ascent were less than those observed in the CMJ, and the chair rise based on range. Nonetheless, joint angle change at the hip was found to be statistically significant. Therefore, to determine these changes had any effect on power, or the change in power during the circuit, a linear regression analysis was conducted. Regression analysis determined that the change in hip angle during the stair ascent task had no significant linear relationship with initial power output in the CMJ ($F_{1,57}$ = 0.534, p=0.468) or on change in power during the CMJ ($F_{1,56}$ = 0.912, p=0.344). Therefore, change in hip angle during the stair ascent had no effect on power during the CMJ or change in power in the CMJ.

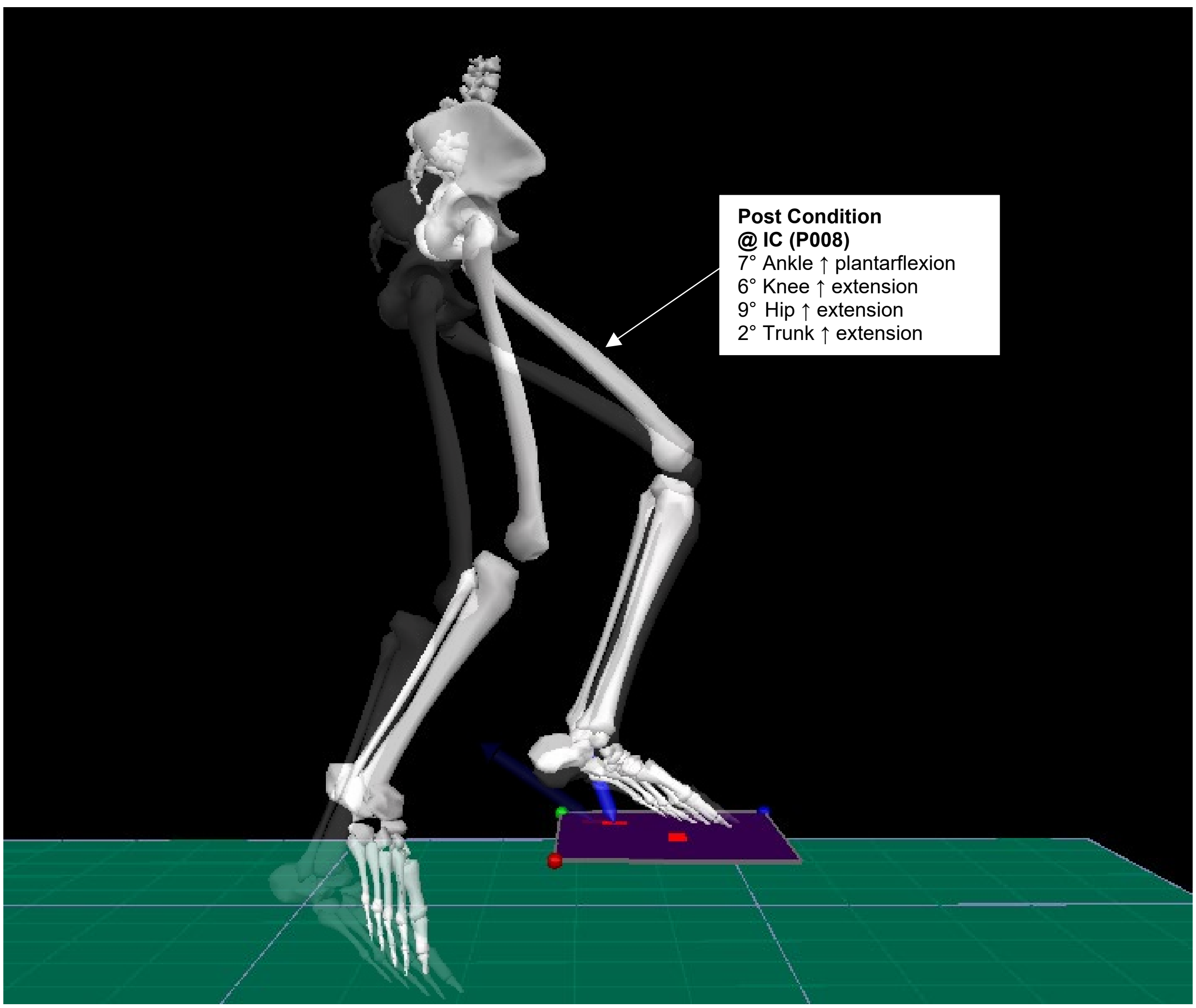

Figure 6.7 - 'Worst-case' pre and post postures during stair ascent at peak IC. The posture in the pre condition is shown as a shadow, while the post condition is shown in the forefront. This participant represented the largest change in joint angle (24°) across four joints observed throughout the circuit. Note that the joint angle changes reported are for the contact limb only.

Finally, during the stair descent task, at the instant of peak ground reaction force, the greatest changes in joint angle from pre to post were observed at the ankle with an RMSD value of 4.30° while the trunk showed the lowest RMSD value of 2.05° (**Table 6.4**).

Table 6.4 – Changes in joint angle in the stair descent at the instant of peak GRF. * Indicates a significant difference from zero (α = 0.05). 'ext' or 'flex' indicate an extension or flexion dominance at post.

	Ankle	**Knee**	**Hip**	**Trunk**
RMSD (°)	4.30	3.27	2.87	2.05
Mean (°)	-1.04 (flex)	-0.55 (ext)	0.85 (ext)	**-1.15 (flex)***
SD (°)	4.29	3.37	3.30	2.24
Min (°)	-15.91	-9.38	-5.25	-6.86
Max (°)	9.05	8.60	10.48	5.64
Range (°)	24.96	17.98	15.73	12.50

Based on mean joint angles, between pre and post during the stair descent, the ankle joint tended towards greater dorsiflexion, the knee joint towards greater extension, the hip joint towards greater extension and the trunk towards greater extension. The change in trunk angle was significantly different from zero with an average of 1.15° (95% CI: -0.57° to -1.72°) more extension in the post condition (see **Figure 6.10** for all task 95% CIs). A Shapiro-Wilk test determined that trunk angle change during stair descent was normally distributed (W=0.961, df=40, p=0.184). The full distribution of all joint angle changes in the stair descent is shown in **Figure 6.8** with greatest outlier participant codes indicated.

Joint angle changes during the stair descent were lesser in range than those of the CMJ but comparable to those of the stair ascent. As well, the trunk showed a statistically significant change from pre to post. To determine if the change in trunk angle had any effect on power or the change in power during the circuit, a linear regression analysis was conducted. Regression analysis determined that change trunk angle during the stair descent had significant linear relationship with initial power output in the CMJ ($F_{1,39}$ = 3.598, p=0.065) or on change in power in the CMJ ($F_{1,38}$ = 0.086, p=0.771). Therefore, the change in trunk angle had no significant effect on power output during the CMJ, or on change in power during the CMJ.

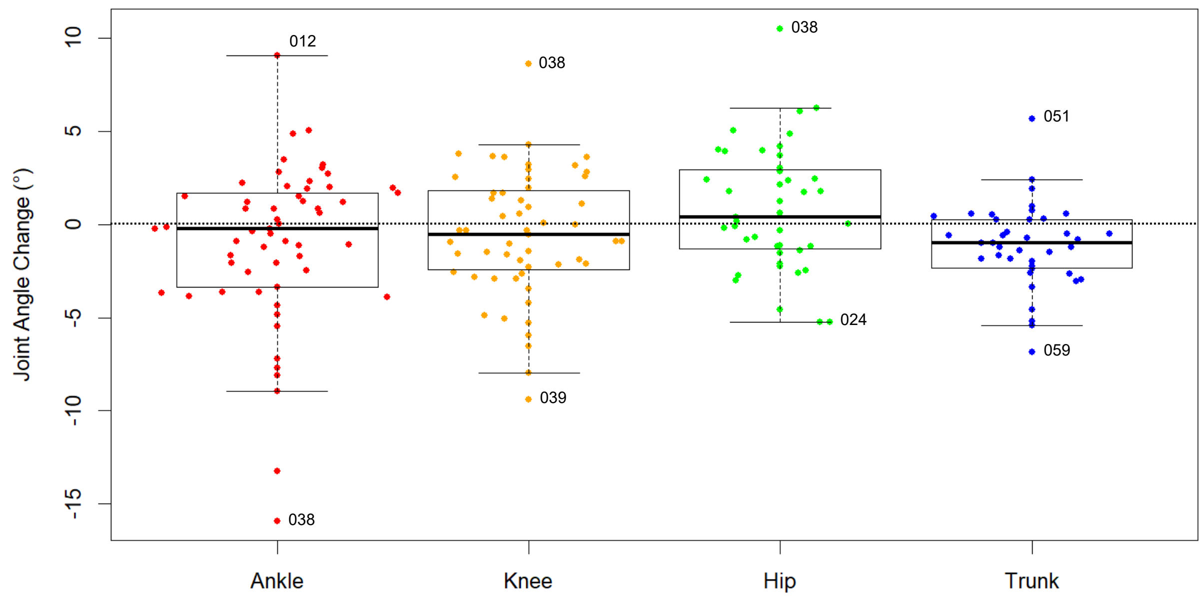

Figure 6.8 - Boxplot of joint angle change from the first two to the last two stair descents (pre to post). On average, between pre and post during the stair descent, the ankle joint tended towards greater dorsiflexion, the knee joint towards greater extension, the hip joint towards greater extension and the trunk towards greater extension. However, the variability between participants was large with all joint-level coefficients of variation greater than 1.96 where 0.5 is typically considered a broad distribution. As well, seven outliers were indicated (within 1.5 to 3.0 IQR), two of which were observed at the ankle, knee and trunk, and one at the hip. A 95% CI suggested that the changes in trunk angle were significantly different from zero (-0.57, -1.72). Data labels show participant code.

The height of the stool and stair riser did not change between participants. This could be perceived as a disadvantage to our participants who were shorter in stature as they would require greater joint angles to complete the same task. At the same time, the standard stool height is more applicable in day-to-day life (Petrella et al., 2005). Nonetheless we investigated the changes in joint angle normalized to body height. When normalized to body height (Δ° / m), the same movement patterns emerged – greater trunk extension in the chair rise at post, greater hip flexion in the stair ascent at post, and greater trunk extension in the stair descent at post. This suggests that on average, changes in kinematics during the circuit were not affected by participant height.

Once again, for context, a screen capture from Visual3D model-space shows the precise pose of the participant (038) who showed the most change between pre and post timepoints across all joints, despite the same peak GRF event trigger (**Figure 6.9**). This change was calculated as the mean of the absolute value of joint angle change per participant (and is therefore independent of direction). This participant showed 35° of joint angle change across four joints – ankle, knee, hip, and trunk – while the average across the whole cohort was 9°.

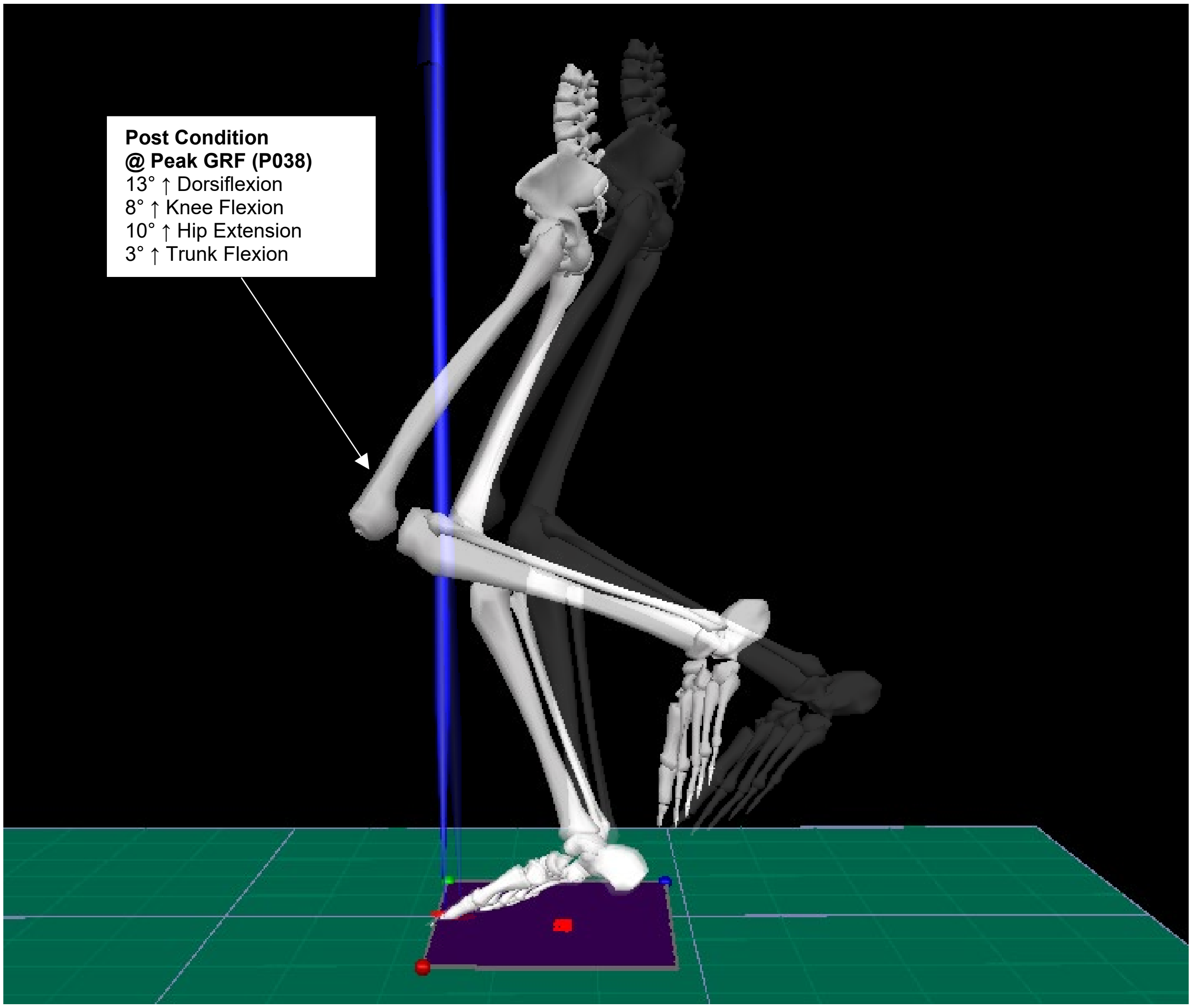

Figure 6.9 - 'Worst-case' pre and post postures during stair descent at peak GRF. The posture in the pre condition is shown as a shadow, while the post condition is shown in the forefront. This participant represented the largest change in joint angle (35°) across four joints observed throughout the circuit. Note that the joint angle changes reported are for the contact limb only.

Overall, the CMJ demonstrated the greatest range of joint angle change, despite not being consistent. More consistent, smaller changes were observed at the trunk in the chair rise, the hips in the stair ascent and the trunk in the stair descent. No joint angle changes during the CMJ, or any statistically significant joint angle changes during the functional tasks could predict initial power in the CMJ or change in power in the CMJ ($p>0.065$). A summary of all confidence intervals is provided in **Figure 6.10**.

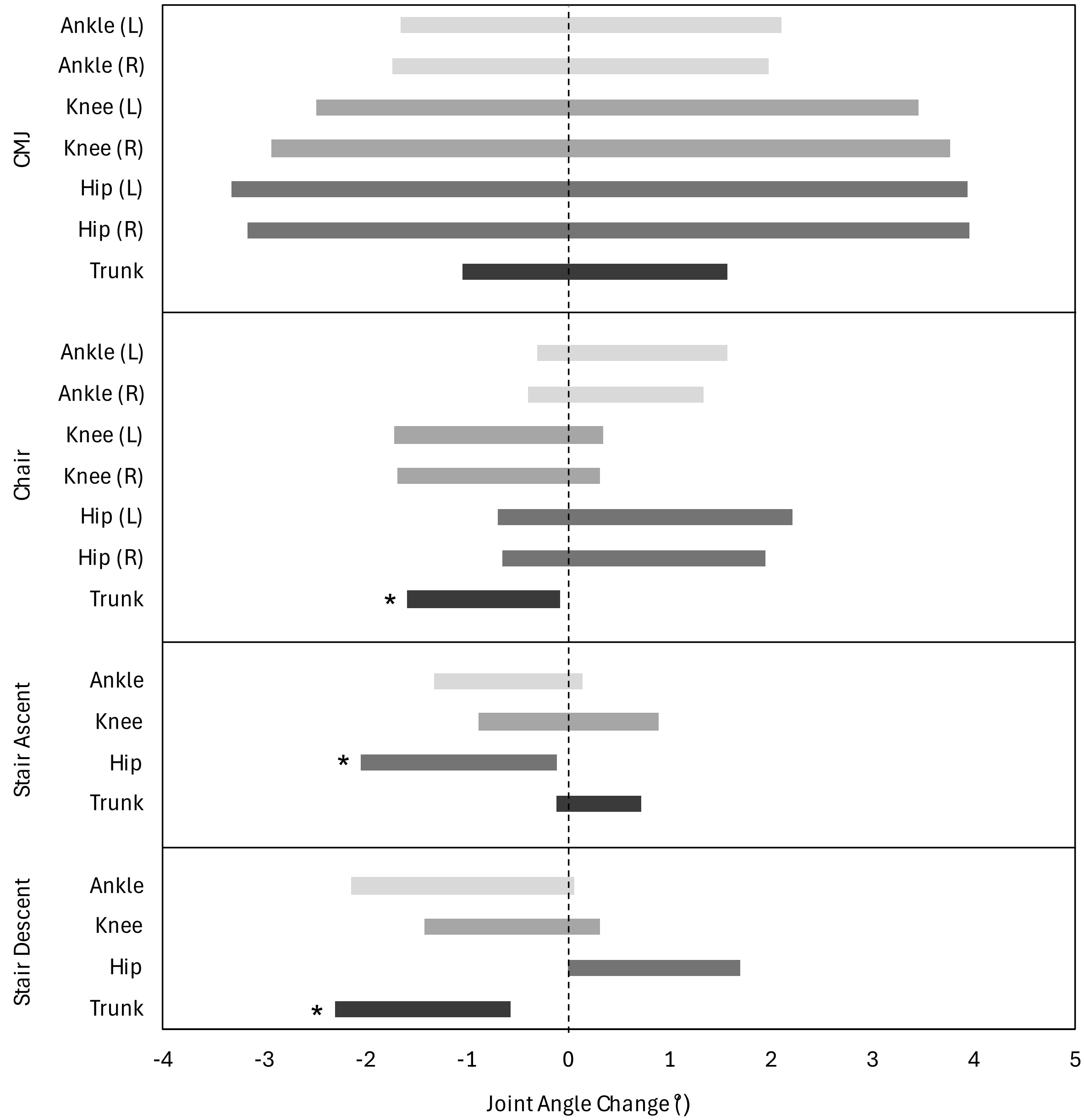

Figure 6.10 - Forest plot showing the 95% confidence intervals of joint angle change at each joint, during each task from pre to post. Trunk angle during chair rise became significantly more extended, hip angle during stair ascent became more flexed, and trunk angle during stair descent became more extended. Shading delineates the various joints. Significant changes are indicated with (*).

Key Findings

- No effect of age, sex or activity level on change in CMJ power from pre to post
- No significant linear relationship between joint angle changes during any task and power or change in power.

- In the CMJ, there was no consistent or statistically significant kinematic change. Changes of approximately ±35° were observed at the knees and hips.
- In the chair rise, there was a statistically significant change to a more extended trunk with joint angle changes of approximately ±10° at all joints except the trunk.
- In the stair ascent, there was a statistically significant change to a more extended hip with joint angle changes of approximately ±10° at the hip.
- In the stair descent, there was a statistically significant change to a more flexed trunk posture with joint angle changes of approximately ±10° at all joints except the trunk.

Discussion

Overview

This section discusses the particular experimental conditions known to elicit differential fatigue in older versus younger adults and compares those conditions to the circuit used in this study. The (null) change in power observed in this study is compared to the results of several other studies that used CMJ protocols taken to exhaustion, albeit in younger adults, with opposing findings. The kinematic results presented above are also compared to existing literature again showing contrary findings – significant kinematic changes were often detected with fatigue. Taken together, this suggests that the circuit was likely not sufficient to induce fatigue.

The Effect of Age

There was no effect of age, sex and activity level on changes in lower-body specific power and thus, we were unable to conclude that older adults were more fatigable than younger adults in the context of our circuit. This finding is in contrast with a 2011 systematic review and meta analysis by Christie and colleagues. In their review, they summarized 60 effects across 37 studies investigating muscle fatigue in young versus older adults with several moderating variables such as contraction intensity (maximal vs submaximal), contraction type (isometric vs dynamic), index of fatigue (endurance time, change in force vs change in power), et cetera. In general, older adults were more fatigable across all moderating variables (effect size of 0.56). However, when looking more closely, they determined that older adults were less fatigable in two conditions – when using dynamic contractions (as opposed to isometric; effect size of -0.12) and when using decreased power as the index of fatigue (as opposed to endurance time or decreased force; effect size of -2.5). In 2018, Kruger and colleagues (2018) published an update to the meta-analysis by Christie and colleagues. In their review they identified six additional papers investigating fatigue measured by decreased power in older adults. Their review agreed with the work of Christie and colleagues in the sense that older adults were less fatigable in isometric fatiguing tasks compared to young adults. They added however, that older adults appear more fatigable when power is used as the index of fatigue and fatigue is induced by dynamic *or* isometric tasks, the latter being a novel finding. More recently, Paris and colleagues (2022) conducted another meta-analysis regarding fatigue in older adults. They added that older adults typically fatigue faster and more assuredly when the dynamic fatiguing task is isotonic. They suggested that the constrained velocity of isokinetic tasks can nullify differences in fatigability between young and old participants. With these meta-analyses in mind, it is surprising that our assessment, which used a dynamic movement (CMJ) with change in power as the index of fatigue, failed to show any appreciable age effects. There are two possible explanations for our null finding.

1. Fatigue was not present, and CMJ power correctly signalled its absence.
2. Fatigue was present, and CMJ power failed to measure it.

In support of the first explanation, it is possible that some features of the circuit, while purposefully designed, may have prevented sufficient development of fatigue. Firstly, the circuit was designed to allow participants of all ages and activity levels to complete several rounds in the 15-minute timespan at their own pace. In line with this goal, the circuit contained 2 rounds of functional tasks bookended by a CMJ to assess fatigue. A more fatiguing circuit could have contained more unbroken rounds of functional tasks, but this would have risked capturing fewer fatigue measurements. Secondly, participants were not time-constrained within each round and were encouraged to rest as needed. A more fatiguing circuit could have constrained participants to 1-minute rounds (for example). However, this would also have risked collecting insufficient data in some participants in addition to a loss of contextual validity, as most daily tasks are not time-bound, and participants would certainly rest at home if they needed to. Lastly, the CMJ was designed to be the index of fatigue, and not the mechanism of fatigue. The functional tasks were intended to induce fatigue which was to be measured by change in power in the CMJ. If the data supported our hypothesis, this prospective result would have been very meaningful as it would have implied that underactive older adults fatigued to a greater extent than younger adults, or active older adults, *simply when doing functional tasks*. A more fatiguing circuit could have involved consecutive CMJ (see Bosco Test references in **Table 6.5**). Older adults would probably have been more likely to fatigue during a Bosco-type circuit, but the results would be less meaningful in the sense that older adults don't rely on consecutive CMJ during daily life. Therefore, making the circuit feasible for participants of all ages and most physical capacities, likely also had the effect of making it too easy for most participants. To underscore this point, it may be helpful to review some more demanding experimental protocols in existing literature, which have successfully developed and measured fatigue using power.

Existing CMJ Protocols

A majority of the CMJ investigations have been conducted in young cohorts however there is one exceptional case provided by Petrella and colleagues (2005). They assessed fatiguability of velocity and power in young and old adults during a knee extension and a rapid sit to stand (STS) task. During the 10-repetition knee extension task (conducted with 40% MVC), older adults decreased their contractile velocity by 25% by the 10th contraction ($p<0.05$) whereas the younger adults showed no significant change ($p=0.08$). During the rapid STS task, all adults showed decreased force (12%), velocity (20%) and power (27%) by the 10th repetition with *no age-differences* in power decline noted. The findings of Petrella and colleagues show that differences in fatigability in older adults, when induced and measured by decreased power, may appear more readily in single joint tasks versus multi-joint tasks such as the CMJ. Nonetheless, the entire cohort displayed decreased power by the end of the test, contrary to our assessment. A separate study in 2021 investigated the effect of a low and high dose jump protocol (7 versus 14 sets of 10 with one minute rest between sets) on jump-based outcomes in 17 males aged approximately 27 (Knihs et al., 2021). They found that power decreased similarly following both a low and high dose of jumps (3.8 versus 4.0W/kg decrease) but that jump height changed to a greater extent following the high dose of jumps (2cm versus 5.7cm). The authors suggested that measures of power are thus more sensitive to lower amounts of fatigue and that jump height was appropriate to measure greater amounts of fatigue. Trinkunas and colleagues (2011) investigated changes in CMJ performance during a 30s Bosco test in 30 sprint versus 23 endurance athletes aged approximately 21 years. They found that sprinter jump height decreased by 24% and power by 29%, whereas endurance jump height decreased by 19% and power by 24%. Pupo and colleagues (2013) evaluated CMJ performance variables including kinematics in a group of male athletes aged approximately 23 years. By 15% of the test (approximately the 4th jump) power had decreased significantly and hip flexion had increased significantly. By 25% of the test (approximately the 7th jump), jump height had also decreased significantly. Near the end of the test, knee flexion had also increased significantly. Sanno and colleagues (2024) evaluated kinetics and kinematics of 300 consecutive vertical jumps in 15 male students aged approximately 25 years. They found that jump height decreased by approximately 11%, and power by 6% accompanied by 5% greater hip flexion. In 2002, Rodacki and colleagues evaluated kinematic changes in the CMJ in 11 male athletes aged 23 years before and after performing a knee extensor exercise to volitional exhaustion. Importantly, they measured ankle, knee and hip angle at the instant of joint extension, following change of direction in the CMJ which is very similar to the peak GRF event used in this analysis [5]. They identified no significant changes to ankle, knee or hip joint angles from pre to post (0.7°, 2.0° and 4.5° respectively). Jump height decreased to 86% of initial, despite observing no change in peak power. These results suggest that kinematic changes may not necessarily be observed

[5] We measured these same joint angles at the instal of peak GRF instead which is roughly coincident with the change of direction. A brief analysis of our data showed that the ankle change of direction occurred 6 frames after peak GRF, the knee change of direction occurred 3 frames after peak GRF and the hip change of direction occurred 3 frames prior to peak GRF. These data are collected at a sample rate of 80Hz thus the difference in our event timing is at most 6 frames, or 0.075 seconds.

with fatigue, although they are typical as we'll see in the remaining literature. McNeal and colleagues (2010) evaluated kinetic and kinematic changes in 11 young males and nine females aged approximately 22 years during a 60-second Bosco test (consecutive hands-on-hips CMJ). By 20 seconds into the test, they observed decreased flight time across the cohort in addition to increased ankle dorsiflexion and knee extension, but no significant change to hip angle.

In general, CMJ tasks performed to fatigue (*in young adults)* results in decreased jump height (McNeal et al., 2010; Rodacki et al., 2002) or both reduced jump height and power (Knihs et al., 2021; Pupo et al., 2013; Sanno et al., 2024; Trinkūnas et al., 2011). Therefore, fatigue in the CMJ can be expected to show statistically significant decreases in power, jump height, or both. Fatigue in the CMJ is also likely to be accompanied by statistically significant changes in joint angle. Based on these facts, it is most likely that fatigue was not present and the positive change in power correctly identified such.

Some literature does also support the second explanation – that fatigue of some sort was present and decreased power during the CMJ failed to measure it. Firstly, a majority of the literature related to fatigue in older adults where power is used as the index of fatigue are conducted with *single joint task*s (Christie et al., 2011; Krüger et al., 2018; Paris et al., 2022). This differs from our protocol which used a complex multi-joint movement as the index of fatigue. Multi-joint movements are less constrained in the sense that kinematics, angular velocity and torque are free to vary across joints, making comparison to data from isolated single joint assessments difficult. Research conducted by Sanno and Colleagues (2024) demonstrated a decrease in lower-body power of 6% after 300 consecutive CMJs, but a greater loss of 17% mean power at the ankle and 15% at the knee. Future analysis will therefore investigate whether changes in joint power, or changes to the distribution of joint powers occurred. There are at least two other instances in the literature where fatigue of some sort was present but no decline in CMJ height was observed. One such study was conducted in a cohort of 25-year-old males and found no difference in CMJ height following a fatiguing running protocol (Gao et al., 2022). The second was evaluated in a cohort of 15 males aged 24 years following a similar fatiguing running protocol. Once again, no difference in jump height between pre and post was found (Yu et al., 2020) despite the volitional exhaustion achieved during running. It appears that running induced fatigue may not fatigue a CMJ to the extent that jumping itself does. This is consistent with the concept of task dependency which dictates that “the dominant mechanism [of fatigue] is specific to those processes that are stressed during the fatiguing exercise” (Enoka and Duchateau, 2008). Our circuit may have generated a similar effect, whereby the impairments developed during the functional tasks were not sufficiently specific to impair the CMJ and thus were not measured. Additional future research considerations are discussed in the limitations section.

Kinetic and Kinematic Changes

It was also possible that changes to movement may have obscured the development and detection of fatigue. Therefore, changes in joint angle were assessed in all four tasks, at pre and post timepoints, at bilateral ankle, knee, hip and at the trunk. In the post condition of the CMJ, there were no joint angle changes that reached statistical significance. All joint changes were less than one degree, on average and the ranges of joint angle change were large with over 70° observed at the left hip – one participant developed 33° greater hip flexion and yet another developed 38° of hip extension. Functionally, these results imply that some participants made large changes to their strategies, and that these changes were inconsistent. As a further example, although it cannot be deciphered directly from **Figure 6.2**, some participants showed greater ankle extension, greater knee flexion, greater hip extension and greater trunk flexion at post (participants 008, 026, 032, 040), while others showed the exact opposite strategy of greater ankle flexion, greater knee extension, greater hip flexion and greater trunk extension at post (004, 007). Regardless, linear regression analysis failed to identify a significant relationship between joint angle change and change in power output. From this we can conclude that the large and inconsistent joint angle changes observed did not impact the change in power output.

Revisiting Hypothesis

3.1 Underactive older adults (>50 years) will show fatigue in the post condition, quantified by reduced power in the CMJ.

Failed to reject the null hypothesis.

Conclusion to RQ #3

No effect for age or activity level was found on changes in lower-body power during the functional task circuit. On the contrary, our whole cohort showed a small mean increase in power throughout the circuit measured by confidence interval. As well, there were no statistically significant kinematic changes in the CMJ, and the changes observed were not related to change in jump height or power. Furthermore, while there were statistically significant kinematic changes observed during the functional tasks, none of the changes showed a significant linear relationship with power or with change in power. Overall, the mean increase in power throughout the circuit and the lack of significant kinematic changes in the CMJ suggest that the circuit was insufficiently fatiguing. Future research should investigate changes in lower-body joint power during functional tasks at pre and post timepoints.

7 Research Question #4: What is the effect of hip and trunk posture on lower body power output?

Introduction

Power production during the CMJ can also be affected by technical considerations such as degree of knee flexion (Moran and Wallace, 2007), hip flexion (Clansey and Lees, 2010) trunk flexion (Vanrenterghem et al., 2008) and the usage of arm swing (Lees et al., 2004). Joint range of motion also changes with age. For example, older adults lose 6-7 degrees of hip flexion per decade starting at approximately age 60 (Stathokostas et al., 2013) and trunk flexion can begin to decline by approximately age 40 in males and females (Intolo et al., 2009). Therefore, it is possible that restricted joint range of motion, particularly at the hip and trunk may have impacted performance in the CMJ. This research question aims to investigate the relationship between hip and trunk postures, age, and power output in the CMJ.

Hypotheses

4.1 Greater trunk flexion angle at peak GRF will be positively associated with lower-body specific power.

4.2 Greater hip flexion angle at peak GRF will be positively associated with lower-body specific power.

Methods

The sample size for Hypothesis 4.1 and 4.2 was 59 (20 females and 39 males). Three females were omitted from an otherwise complete dataset due to motion capture technical difficulties. The protocol for Hypothesis 4.1 and 4.2 consisted of three maximal CMJs performed with motion capture and force plates, facilitating the calculation of hip and trunk postures at the instant of peak GRF. A neutral reference position was used to calculate the change in joint position. Hypotheses 4.1 and 4.1 were analyzed via multiple linear regression. Additional analyses were conducted with univariate ANOVA and was presented graphically with histograms. More detailed methods can be found in the common methods section.

Results

Analysis of trunk flexion angle at the instant of peak GRF (**Figure 7.1**, left) and peak power production (**Figure 7.1**, right) suggests that greater trunk flexion angle is positively associated with greater lower-body specific power production, as indicated by darker green frequency counts in the range of -20° to -30° trunk flexion.

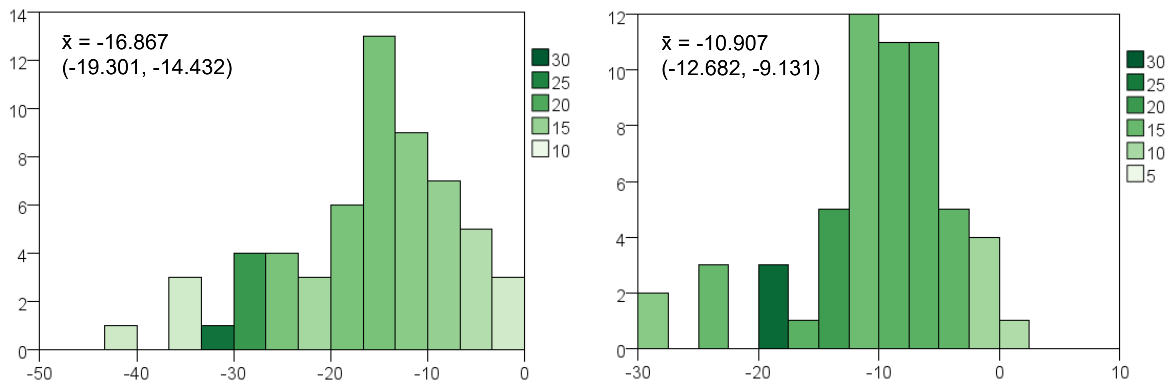

Figure 7.1 - The relationship between trunk flexion and peak lower-body power during the CMJ at the instant of peak GRF production (left) and peak power production (right). Based on this representation, greater trunk flexion during the CMJ appears to be associated with greater lower-body power output. Color scale shows lower-body specific power. Data labels at top left show mean and 95% CI of trunk angle. Further analysis by age-group provides additional detail.

A multiple linear regression analysis investigating the effect of age, trunk flexion angle at peak GRF, and trunk flexion angle at peak power on lower-body specific power revealed a statistically significant model ($F_{3,57}$ = 16.526, R^2 = 0.692, p<0.001) with significant contributions from age (r = -0.661, β = -0.299, p<0.001; Hypothesis 4.1) and trunk flexion angle *at peak GRF* (r = -0.130, β = 0.310, p=0.050). Trunk flexion angle *at peak power* did not contribute significantly to the model (p=0.180). Additional age-group analysis via one-way ANOVA ($F_{3,58}$ = 6.035, p<0.001) revealed that adults aged >70 demonstrated significantly less trunk flexion at the instant of peak GRF production compared to adults aged <30 years (Δ = 14.028°, p<0.001). **Figure 7.2** and **Figure 7.3** below show the extent of trunk flexion used by participants in the various age groups at the instant of peak GRF and peak power production.

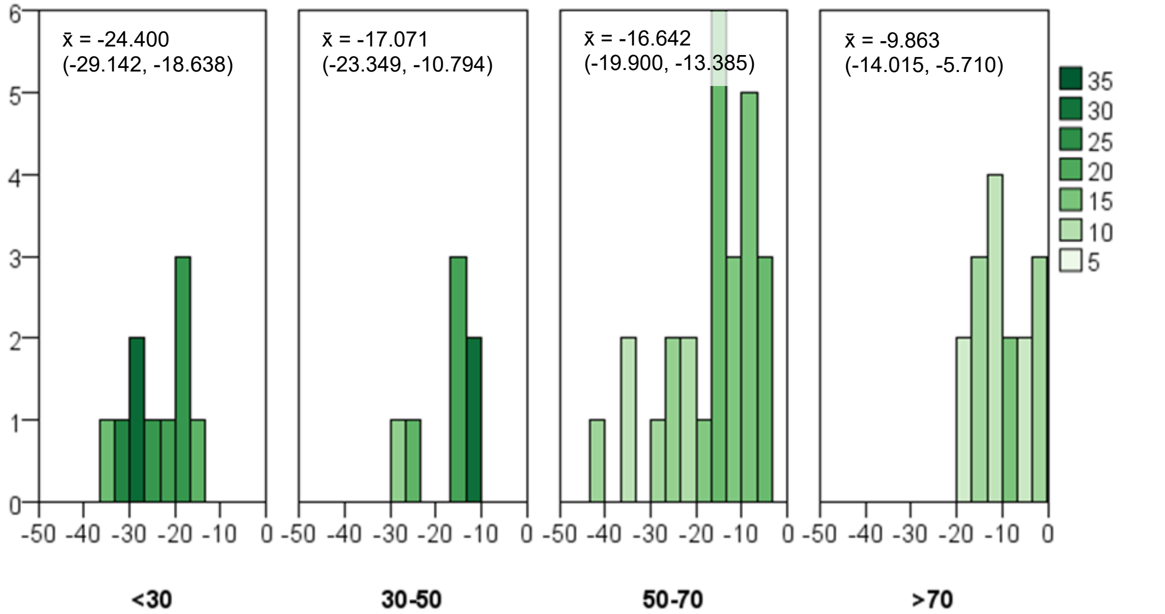

Figure 7.2 – The relationship between trunk flexion and peak lower-body power at the instant of peak GRF, organized by age-group. A one-way ANOVA ($F_{3,58}$ = 6.035, p<0.001) determined that participants aged less than 30 years showed significantly greater trunk flexion than those aged greater than 70 years of age at the instant of peak GRF (Δ = 14.028°, p<0.001). Data shown represent the difference in sagittal plane trunk segment angle between a neutral standing position and the position adopted during the CMJ at the instant of peak GRF production. Panels show age groups, x-axis shows degrees (negative = trunk flexion), y-axis shows frequency counts. Labels at top show mean and 95% CI of trunk flexion angles. Color scale shows lower-body muscular power output (W/kg).

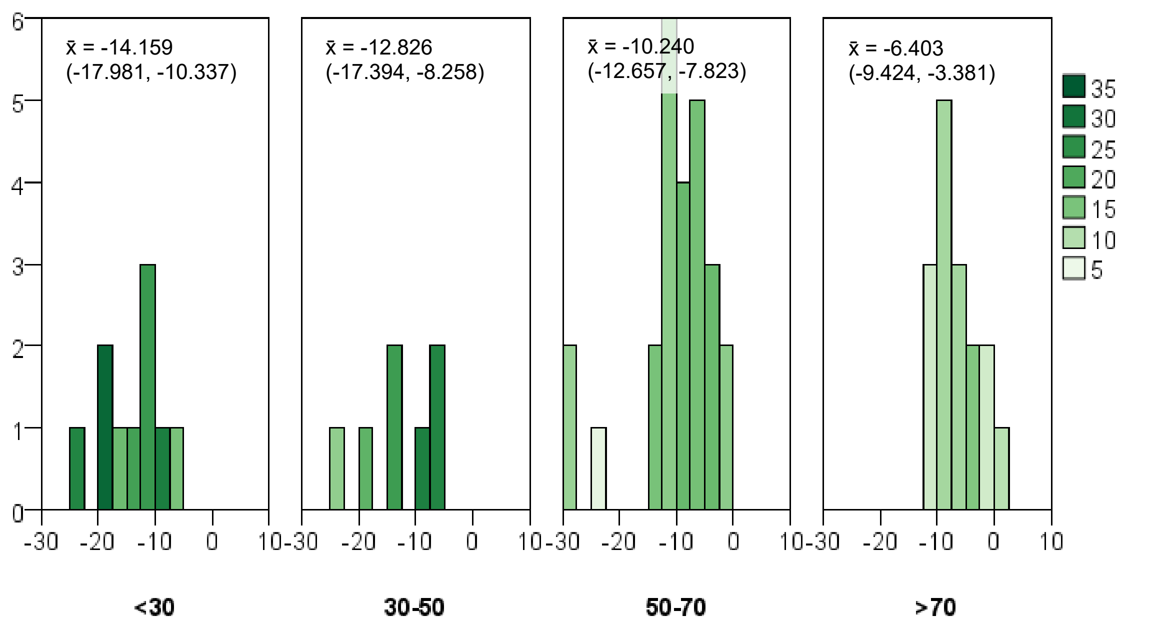

Figure 7.3 – The relationship between trunk flexion and peak lower-body power at the instant of peak power, organized by age-group. A one-way ANOVA ($F_{3,57}$ = 3.997, p=0.012) determined that participants aged less than 30 years showed significantly greater trunk flexion than those aged greater than 70 years of age at the instant of peak power (Δ = 7.765°, p=0.014). Data shown represent the difference in sagittal plane trunk segment angle between a neutral standing position and the position adopted during the CMJ at the instant of peak GRF production. Panels show age groups, x-axis shows degrees (negative = trunk flexion), y-axis shows frequency counts. Labels at top show mean and 95% CI of trunk flexion angles. Color scale shows lower-body specific power output (W/kg).

Analysis of hip flexion angle at the instant of peak GRF (**Figure 7.4**, left) and peak lower-body power (**Figure 7.4**, right) suggests that greater hip flexion angle is also associated with greater lower-body specific power production, as evidenced by the darker blue frequency counts in the range of 100° hip flexion at the instant of peak GRF production (left; Hypothesis 4.2). The relationship between hip flexion at the instant of peak power production (right) is less clear.

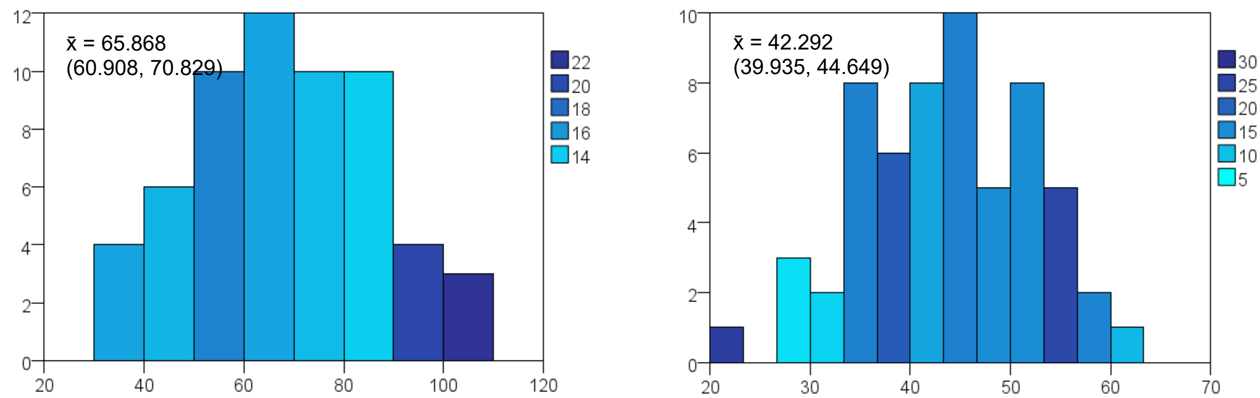

Figure 7.4 - The relationship between hip flexion and peak lower-body power during the CMJ at the instant of peak GRF production (left) and peak power production (right). Based on this representation, greater hip flexion during the CMJ appears to be associated with greater lower-body power output, especially at peak GRF (left). Color scale shows lower-body specific power. Data labels at top left show mean and 95% CI of hip angle. Further analysis by age-group provides additional detail.

A multiple linear regression analysis investigating the effect of age, hip flexion angle at peak GRF and hip flexion angle at peak power on lower-body power, revealed a statistically significant model ($F_{3,57}$ = 16.331, R^2 = 0.471, p<0.001) with significant contributions from age only (r = -0.662, β = -0.284, p<0.001; See Research Questions #1 & #2). Here we found that hip angle at peak GRF and at peak power did not make statistically significant contributions to power production during the CMJ (p>0.076). Additional categorical analysis via one-way ANOVA ($F_{3,57}$ = 3.898, p=0.046) revealed that participants aged greater than 70 displayed 14.847° less hip flexion compared to those aged 50-70 years at peak GRF and 8.193° less hip flexion at the instant of peak power as well ($F_{3,57}$ = 4.085, p=0.013). However, according to the regression model, these postural differences did not appear to impact power output. **Figure 7.5** and **Figure 7.6** below show the extent of hip flexion used by participants in the various age groups at the instant of peak GRF, and peak power production.

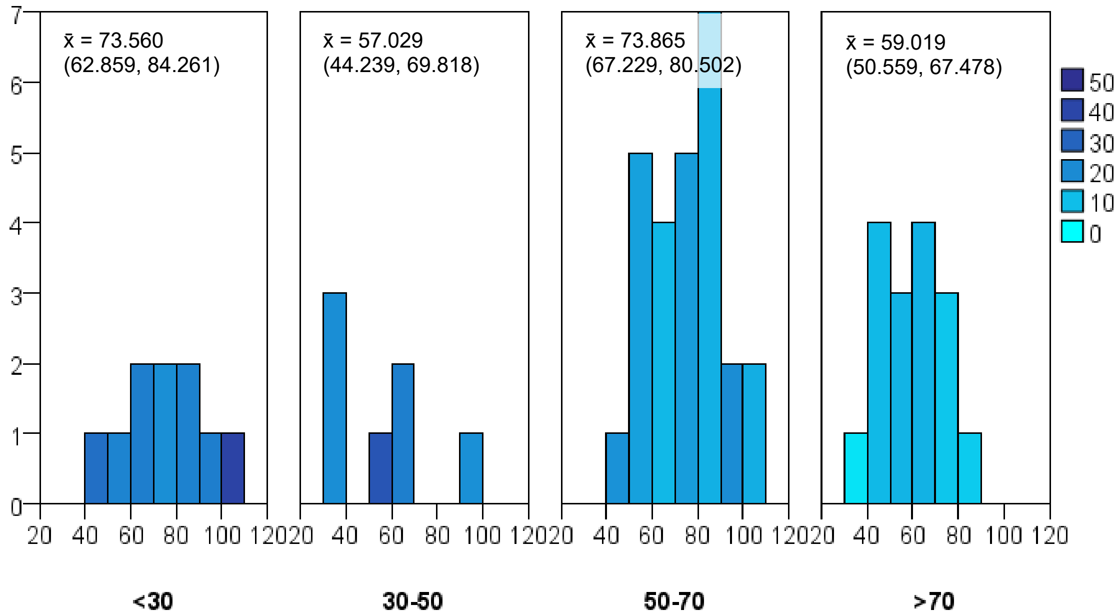

Figure 7.5 – The relationship between hip flexion and peak lower-body power at the instant of peak GRF, organized by age-group. At the instant of peak GRF production during the CMJ, participants aged greater than 70 displayed less hip flexion compared to those aged 50-70 years ($F_{3,57}$ = 3.898, Δ = 14.847°, p=0.046) however these differences were not predictive of power (p>0.076). Data shown represent the difference in sagittal plane lumbar segment angle between a neutral standing position and the position adopted during the CMJ at the instant of peak GRF production. Panels show age groups, x-axis shows degrees (positive = hip flexion), y-axis shows frequency counts. Labels at top show mean and 95% CI of hip flexion angles. Color scale shows lower-body muscular power output (W/kg).

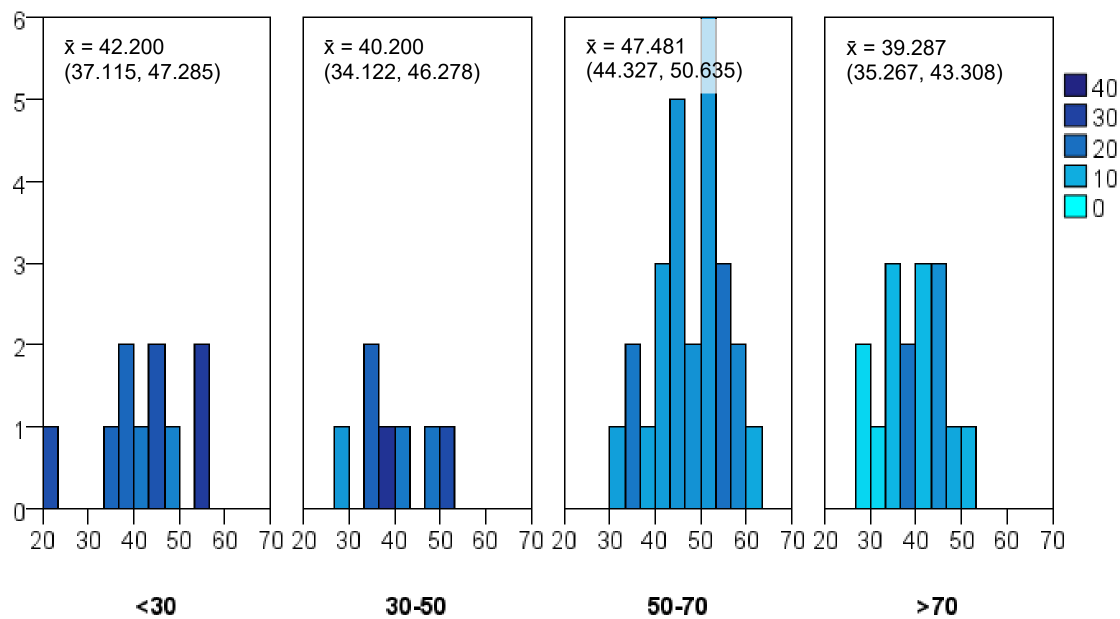

Figure 7.6 – The relationship between hip flexion and peak lower-body power at the instant of peak power, organized by age-group. At the instant of peak power production during the CMJ, participants aged greater than 70 displayed less hip flexion compared to those aged 50-70 years ($F_{3,57}$ = 4.085, Δ = 8.193°, p=0.013) however these changes were not predictive of greater power (p>0.076). Data shown represent the difference in sagittal plane lumbar segment angle between a neutral standing position and the position adopted during the CMJ at the instant of peak lower-body power production. Panels show age groups, x-axis shows degrees (positive = hip flexion), y-axis shows frequency counts. Labels at top show mean and 95% CI of hip flexion angles. Color scale shows lower-body muscular power output (W/kg).

These findings suggest that older adults tended to use less hip and trunk flexion compared to the younger adults. The greater trunk flexion in younger adults appeared to make a meaningful contribution to CMJ performance, with gains of approximately 0.3W/kg for every degree of trunk flexion predicted by our regression model. Thus, one possible mechanism for reduced power output in older adults appears to be reduced trunk flexion. Whether or not this mechanism is causative remains to be investigated.

Key Findings

- Lower age and greater trunk flexion at peak GRF predict higher power during CMJ.
    - Approximately 0.3W/kg gained per degree of trunk flexion ($r = 0.130$) and 0.3W/kg lost per year of age ($r = 0.661$).
- Adults aged >70 display less trunk flexion than adults aged <30 at peak GRF and peak power ($p<0.014$), with above mentioned consequences on power.
- Adults aged 70+ display less hip flexion than adults aged 50-70 at both time points ($p<0.046$), with no clear consequences on power observed.

Discussion

Overview

This section discusses some potential theories related to trunk angle during the CMJ, which was a strategy that was uniquely adopted by younger adults in this study. Some technical reasons are offered to explain the lack of relationship between hip angle and power output in this study. As well, the technical nature of the CMJ as it relates to measuring power in older adults is discussed.

The Effect of Trunk Angle

Regression analysis found that increased trunk angle predicted higher power output in the CMJ, and this strategy was adopted primarily by younger adults aged <30 years. Existing literature has demonstrated that trunk angle can influence jump height. For example, constraining trunk angle to a more upright posture (approx. 21° of inclination versus approx. 44° unconstrained) during a CMJ leads to a 10% decrease in jump height. These effects are thought to be mediated by decreased hip power and a concomitant increase in knee power during the CMJ (Vanrenterghem et al., 2008). As well, the effect of age on power output is well documented (including here in Research Questions 1 & 2). However, the strategy of greater trunk flexion in predominantly younger adults is a novel finding and invites several theories, the first of which is related to vision and balance. A recent review showed that older adults are more reliant on visual feedback compared to young adults for postural control and balance (Osoba et al., 2019). Thus, it is possible that keeping one's trunk more vertically oriented may help to more easily maintain visual input for balance in the older adult.

The second theory relates to centre of pressure (COP). The dominant vertical rise strategy is a documented maladaptive tactic used by older adults in the chair rise task (Scarborough et al., 2007). The participant's centre of mass was kept posterior to the ankle joint for the majority of the chair rise task. It was theorized that this strategy was adopted due to fear of dizziness or of falling forward (Scarborough et al., 2007). Evidence also suggests that older adults show increased centre of pressure (COP) excursion and decreased margin of stability (MOS) during quiet standing and balance tasks (Matson and Schinkel-Ivy, 2020). However, more challenging volitional functional tasks, such as a sit-to-stand to gait initiation task, or a lifting-lowering task showed the opposite effect in older adults – reduced COP excursion and increased MOS. The authors suggested that the more conservative balance measures may result from older adults showing greater caution during more challenging functional tasks. Lastly, trunk flexion ROM has been shown to begin to decline with age in both males and females. Between 20 to 70 years of age, lumbar flexion ROM reduced in females by 9.0 degrees and 16.3 degrees in males (Intolo et al., 2009).

Taken together, this existing research suggests that older adults in our cohort were less likely to show a great amount of trunk flexion during the CMJ task potentially to maintain stronger visual input during the jump, to keep the centre of mass more posterior for fear of falling, to generally reduce the COP excursion out of caution, or simply due to a lack of ROM in the lumbar spine.

Our results have important implications on the measurement of power output via the CMJ in older adults. Older adults appear to adopt a more vertical trunk strategy during the CMJ for balance-related purposes. This strategy may limit power output in the CMJ and therefore some age-related losses in power output may be technical in nature, albeit still related to aging via balance deficits.

The Effect of Hip Angle

Regression analysis found that only age was a significant predictor of lower-body power. Hip angle at peak GRF and peak power both failed to make statistically significant contributions to the model. With respect to age being a significant predictor of power, as above, this finding is consistent with the earlier age-related research questions (#1 & #2) showing a strong and consistent negative age-effect on lower-body power. However, the lack of hip angle effect is in contrast with some existing literature. A small, underpowered CMJ study, conducted in six healthy males aged approximately 23 years, showed a moderately strong relationship between peak hip flexion angle and CMJ height (R^2=0.346, p-value not provided), with the highest jumps observed in the participants who showed the greatest hip angles (100 to 130°; Clansey and Lees, 2010). The work of Clansey and Lees measured peak hip angle and jump height, whereas we reported hip angle at peak GRF and corresponding peak power. As well, their analysis used a second order polynomial as opposed to a linear regression line. In 2004, Vanreterghem and colleagues sought to understand how kinematics change during CMJs of increasing effort (Vanrenterghem et al., 2004). In their study of 10 male volleyball players aged approximately 23 years, they found that increased jump height was primarily achieved by increased countermovement depth which was mediated by greater hip flexion. Thus, they found that peak hip flexion amplitude was strongly related to jump height (R^2 = 0.823, p-value not provided). Our data did not support a relationship between hip angle at the instant of peak GRF or peak power production despite the findings of both abovementioned studies. It appears that the relationship between certain kinematic measures and related power output are sensitive, making comparison to existing data challenging.

Revisiting Hypotheses

4.1 Greater trunk flexion angle at peak GRF will be positively associated with lower-body specific power.

Accepted.

4.2 Greater hip flexion angle at peak GRF will be positively associated with lower-body specific power.

Rejected.

Conclusion to RQ #4

Trunk angle, measured at the instant of peak GRF was found to predict lower-body specific power in the CMJ. While the effect of trunk flexion on jump performance has been demonstrated in the literature, a novel addition is that predominantly younger adults were found to use this strategy, whereas older adults did not. Thus, while some of the changes in CMJ performance can be attributed to age, it appears that movement strategies can affect performance as well. Future research should attempt to quantify the effect of differing movement strategies by comparing power output produced on a simple sledge ergometer (or similar) to that of a CMJ, in the same cohort. By comparing these two modalities, it may be possible to understand how much loss of power is driven by physiology versus technical considerations of the CMJ.

8 Research Question #5: Can MRI-diffusion parameters predict muscular power output?

Introduction

The capacity to produce muscular power in humans depends on several factors including some of those assessed in the preceding research questions – age, sex, activity level, and athletic discipline. These factors exert some of their effects through several well-known physiological characteristics of the motor unit: muscle fiber size, pennation angle, the number of parallel cross-bridges, the force per unit cross-bridge, and fiber type, to name a few. Fiber type composition is especially important as the percentage of fast twitch fibers is strongly correlated with peak power (Fitts et al., 1991). Quantifying muscle fiber type in our cohort would likely reveal another strong factor in predicting power output across the lifespan, however, direct measurement of muscle fiber type typically requires a muscle biopsy which is invasive (Baguet et al., 2011) and likely would have impaired recruitment. At the same time, quantifying muscle fiber type might offer a more fundamental understanding of the mechanism through which age, activity level and athletic discipline affect muscular power. For that reason, a novel, MRI-based method, diffusion tensor imaging (DTI) was selected to approximate muscle fiber type in our cohort. This method measures the movement of fluid through cylindrical muscle cells. This fluid movement, typically characterized by fractional anisotropy (FA; a quantitative MRI parameter), was theorized to behave uniquely in type- I muscle fibers allowing for their quantification (Scheel et al., 2013). Similar relationships have been found between FA and maximum muscle power output at the ankle (Scheel et al., 2013a). These two findings form the basis for the research questions presented below.

Hypotheses

5.1 Fractional anisotropy (FA) of the soleus, lateral gastrocnemius and medial gastrocnemius muscle would be inversely proportional to ankle power output.

5.2 Short distance athletes will show lower FA in the plantarflexors compared to both non-athletes and long-distance athletes.

5.3 There will be a linearly decreasing correlation between the proportion of type I muscle fiber in each muscle and the measured FA for that muscle.

Additional Methods

Participants

A total of 36 participants participated in the MRI portion of the study, including 20 athletes aged approximately 21 to 83 years and 16 non-athletes aged approximately 28 to 85 years. For Hypothesis 5.1, when analyzing FA of the soleus muscle, three outlier data points were identified via boxplot and after investigation, they were omitted. Likewise, one outlier was omitted from lateral gastrocnemius, none from medial gastrocnemius and one from tibialis anterior. 'Other' athlete data was withheld from Hypothesis 5.2 due to small sample size, lower mean age and age range. As a result, the sample size for this question was 31. The same participants from Hypothesis 5.1 were included in Hypothesis 5.3 however 5.3 also included 36 data points for extensor digitorum longus FA and 34 participants for fibularis longus FA. A summary of the participants for Research Question #5 is provided below in **Table 8.1**.

Table 8.1 - Description of participants who completed the MRI portion of the study. Exclusions are discussed above.

	Overall			
Sample Size	36			
Age	57.8			
Age Range	23.6 to 85.2			
	Short-Distance	**Long-Distance**	**Non-Athlete**	**Other Athlete**
Sample Size	7	8	16	5
Age	64.9	59.1	56.1	51.4
Age Range	21.5 to 83.6	23.6 to 80.8	28.7 to 85.2	23.6 to 61.2

Measures

Magnetic resonance (MR) imaging was conducted with two distinct sequences serving distinct research questions. The first sequence employed Dixon imaging which resulted in fat and water separated images, facilitating the quantification of muscle cross-sectional area, subcutaneous fat, intermuscular fat and intramuscular fat for all identified muscles. The second sequence consisted of diffusion tensor imaging (DTI) at the shank to non-invasively quantify muscle fiber type, to potentially explain differences in power output which could not be explained by the other factors considered here namely age, sex and PA. Pilot study of thigh muscle DTI revealed significant artifact from breathing and peristalsis. Therefore, the shank region was selected to minimize noise.

The purpose of performing MRI was three-fold. It allowed us to compare muscle mass and fat infiltration across age, but also across activity level and type. Secondly, MRI muscle cross-sectional areas (CSA) were explored as an alternate method to normalize power and joint moment data (RQ #1). Normalizing power output to lean thigh volume (not CSA) has previously been shown to lessen the difference in specific power between younger and older active males (Martin et al., 2000). The final purpose was to perform non-invasive muscle fiber type and/or muscle power assessment via DTI (Scheel et al., 2013b, 2013a). This is important as several studies have noted a dissociation between loss of muscle and loss of performance (Reid et al., 2012). As well, information related to muscle fiber type may help explain these discrepancies in muscle power output.

All imaging was conducted by Manoj Kumar, York University’s MRI technician, on a Siemens MAGNETOM 3T PrismaFit scanner. MR imaging began with a three-plane localizer to locate the region of interest. Following each localizer, DTI and Dixon scans were conducted as described below in **Table 8.2**.

Table 8.2 - MRI sequence parameters. DTI parameters established through pilot testing with guidance from Oudeman and colleagues (2016) and in correspondence with quantitative MRI expert Martijn Froeling.

	Shank Imaging		**Thigh Imaging**	**Hip Imaging**	**Spine Imaging**
Location	1/3 dist. between fib. and lat. mal.		Mid-thigh	Head of GT	L4/L5 disc
Methodology	DTI	Dixon	Dixon	Dixon	Dixon
Imaging plane	Axial	Axial	Axial	Axial	Axial
Matrix (px)	108 x 62	192 x 230	256 x 306	256 x 306	256 x 306
Voxel size (mm)	2.8 x 2.8 x 5	1.4 x 1.4 x 5	1.0 x 1.0 x 1.0	1.3 x 1.3 x 1.0	1.3 x 1.3 x 1.0
Slices	105	44	64	104	104
FoV (mm)	299 x 172	260 x 311	259 x 310	339 x 406	339 x 406
Slice gap	0	0	0	0	0
TR (ms)	3300	6.72	6.72	7.8	7.8
TE (ms)	65	2.46	2.46	2.46	2.46
ETL	36	2	2	2	2
Scan time (m:s)	8:30	1:00	2:00	3:45	3:45
b-values	600	n/a	n/a	n/a	n/a

Procedure

The second lab visit began with a review of the MRI screening form led by York University's MRI technologist and safety officer in the MRI Facility waiting area (zone 2). Once the participant was cleared for an MRI, they entered the MRI control room (zone 3) where the participant changed into medical scrubs. Imaging locations were then identified on the participant using vitamin E capsules. These capsules appear as bright ovals in localizer images, which allows the technologist to centre the field of view over consistent anatomical region in all participants. Four regions of the body were imaged: (1) shank, (2) thigh, (3) hip and (4) spine. The specific procedure for each region is provided next.

1) *Shank imaging*: The distance was measured between the fibular head and lateral malleolus of the ankle. The vitamin-E capsule was placed on the lateral aspect of the shank, at one-third of this distance moving distally from the fibular head (Davison et al., 2017). Four primer movers for ankle plantarflexion and dorsiflexion were targeted: the medial and lateral gastrocnemius, the soleus muscle, and the tibialis anterior.

2) *Thigh imaging*: The distance was measured between the greater trochanter (described below) and the lateral epicondyle of the femur. The vitamin-E capsule was placed on the lateral aspect of the thigh, at the mid-point of this distance (M. V. Franchi et al., 2018). Eight knee extension and flexion prime movers from the quadriceps and hamstrings group were

targeted: rectus femoris, vastus lateralis, vastus intermedius, vastus medialis, biceps femoris long and short head, semitendinosus, and semimembranosus.

3) *Hip imaging*: The head of the greater trochanter was located via manual palpation and identified with a vitamin-E capsule. Imaging of the gluteus maximus musculature was done at an axial plane intersecting the superior tip of the greater trochanter (Perraton et al., 2022).

4) Spine imaging: Imaging was centered at the L4/L5 disc which was located during the localizer scanning procedure. Imaging of the psoas muscle was performed at the midplane of the L4-L5 disc (Arbanas et al., 2013). The gluteus max and psoas are collectively the prime movers for hip extension and flexion.

All distances were measured using a seamstress tape, and all capsules were affixed using one-inch 3M™ Transpore™ medical tape. The participant was then led into the magnet room (zone 4) where they lay in the supine, feet-first position on the MRI table. This position was maintained for shank, thigh and spine imaging whereas hip imaging was done in the prone feet-first position. Hearing protection and the emergency squeeze bulb was provided to the participant, and the bulb was tested. For the supine images, triangular foam blocks were placed under the knee to allow the shank and thigh to maintain their natural form. An 18-channel body coil was centered above the location of interest (typically identified by the vitamin-E capsule, as above) and a 32-channel coil embedded in the table was also active (Siemens BioMatrix Spine 32). Additional foam pads were used to prevent skin contact with the magnet, and/or body coil, wherever used. Thin cotton sheets were used as needed to maintain comfortable body temperature during the procedure. Participants were asked to limit movement during the procedure, especially when the magnet was making noise, which indicates that images are being collected.

Processing

MRI Dixon image analysis was conducted in ITK-SNAP (v4.0.2) which is an open-source multi-platform application used to segment medical images in three-dimensions (see itksnap.org). Analysis began by importing the Water images into the program. Subsequently, Fat images were then imported into the program as additional and separate (as opposed to additional and semi-transparent overlay). This permitted Water and Fat images to be viewed separately, which allowed raters to view and confirm various structures as needed. At this point, the rater imported the relevant label descriptions which were created in advance based on the anatomy known to intersect in each region of interest (**Table 8.3** below). For shank, thigh and spine, all muscles, including those not required for analysis, were measured. Identifying these borders helped identify the borders of the prime movers.

Table 8.3 - List of muscles imaged by MRI.

Shank segments	Thigh segments	Hip segments	Lumbar Spine segments
1 Tibialis anterior	23 Rectus femoris	54 Gluteus Maximus	59 Erector spinae
2 Tibialis anterior fat	24 Rectus femoris fat	55 Gluteus Maximus Fat	60 Erector spinae fat
3 Extensor digitorum longus	25 Sartorius	56 Femur	61 Multifidus
4 Extensor digitorum longus fat	26 Sartorius fat	57 Subcutaneous fat	62 Multifidus fat
5 Fibularis longus	27 Vastus medialis	58 Intermuscular fat	63 Psoas
6 Fibularis longus fat	28 Vastus medialis fat	101 Hip CSA	64 Psoas fat
7 Flexor hallucis longus	31 Adductor longus		65 Quadratus lumborum
8 Flexor hallucis longus fat	32 Adductor longus fat		66 Quadratus lumborum fat
9 Lateral gastrocnemius	33 Gracilis		67 Internal oblique
10 Lateral gastrocnemius fat	34 Gracilis fat		68 Internal oblique fat
11 Medial gastrocnemius	35 Adductor magnus		69 External oblique
12 Medial gastrocnemius fat	36 Adductor magnus fat		70 External oblique fat
13 Soleus	37 Semimembranosus		71 Transverse abdominus
14 Soleus fat	38 Semimembranosus fat		72 Transverse abdominus fat
15 Flexor digitorum longus	39 Semitendinosus		73 Latissimus dorsi
16 Flexor digitorum longus fat	40 Semitendinosus fat		74 Transverse abdominus fat
17 Tibialis posterior	41 Biceps femoris long		75 Rectus abdominus
18 Tibialis posterior fat	42 Biceps femoris long fat		76 Rectus abdominus fat
19 Tibia	43 Biceps femoris short		77 L4/L5 IVD
20 Fibula	44 Biceps femoris short fat		78 Subcutaneous fat
21 Subcutaneous fat	45 Illiotibial tract		79 Deep subcutaneous fat
22 Intermuscular fat	46 Illiotibial tract fat		80 Intermuscular fat
99 Shank CSA	47 Vastus lateralis		81 Visceral fat
	48 Vastus lateralis fat		102 Trunk CSA
	49 Vastus intermedius		
	50 Vastus intermedius fat		
	51 Femur		
	52 Subcutaneous fat		
	53 Intermuscular fat		
	100 Thigh CSA		

The central slice was identified by locating the centre of the vitamin-E capsule in the image. One image above and below this central image was also quantified to minimize type II error. All muscle and subcutaneous fat cross-sectional areas were segmented manually with the polygon utility. Intermuscular and intramuscular fat were more sporadic and as such, were manually segmented with the round paintbrush utility, set to a brush size of between one and three pixels.

The use of signal intensity to objectively identify intramuscular fat has been validated in the paraspinal (D'hooge et al., 2012) and thigh muscles (Engelke et al., 2022). At the time of this writing, it was unclear whether there existed a proven method for segmenting intramuscular and intermuscular fat using signal intensity (Ogawa et al., 2017), especially in a sample with participants aged 22 to 85. For that reason, the analysis in this dissertation was performed manually.

Segmenting was performed in a particular order (**Figure 8.1**). First, the whole cross-sectional area (i.e., complete thigh) was quantified by tracing around the entire slice. Secondly, the subcutaneous fat layer was quantified as the layer of fat deep to the skin but superficial to the superficial fascia. In hip and spine imaging, deep subcutaneous fat was also quantified as the layer of fat deep to the superficial fascia but superficial to the deep fascia. Thirdly, the muscles of interest and any bone were segmented. Fourth, any fat deep to the superficial fascia and outside of a muscle was considered intermuscular fat. Lastly, any fat inside of a muscle belly was considered intermuscular fat.

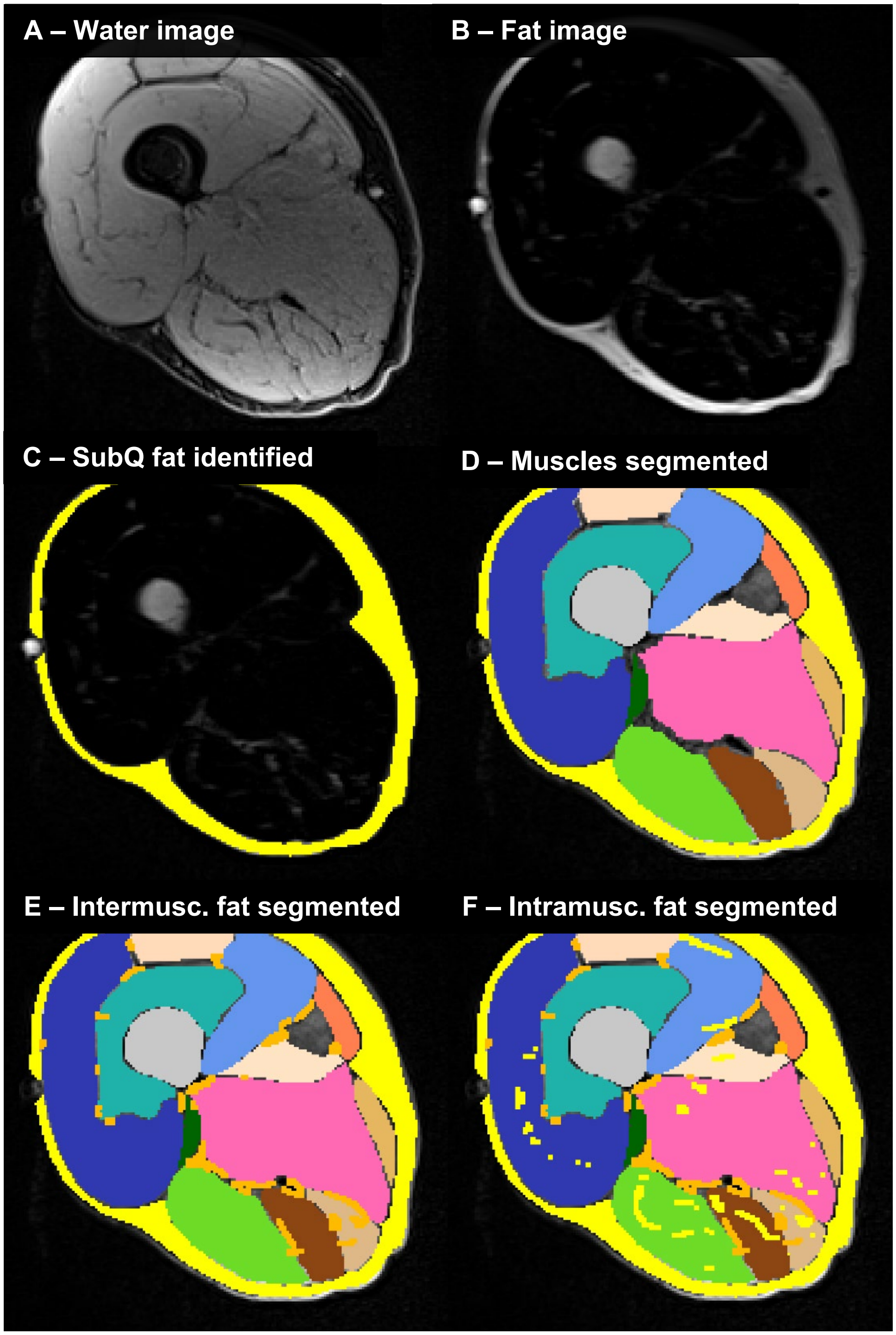

Figure 8.1 – Segmentation of a thigh MR image in ITK-SNAP. Images A and B show the Water and Fat separated images respectively resulting from the Dixon sequence. Image C shows subcutaneous fat segmentation completed, with this layer beginning deep to the skin surface, progressing deep to the superficial fascia lata. Image D shows muscle (and femur) segmentation completed. Image E and F show intermuscular fat and intramuscular fat segmentation completed. The remaining unsegmented space is that of the femoral artery and vein.

MR DTI imaging parameters are provided above in **Table 8.3** and b-values per Ouderman (2016). MRI diffusion tensors were calculated using DTITools, which is a custom toolbox built by Martijn Froeling within Wolfram Mathematica 13.3 (DTI Tools, https://github.com/mfroeling/DTITools). The Wolfram Mathematic pipeline was created by Diana Gorbet, York University's MRI specialist.

DTI processing is highly systematic, consisting of converting image formats, importing images, image superposition, denoising, motion and eddy current correction, perfusion correction followed by calculation of DTI metrics. Data are plotted in Mathematica following each step in the processing sequence to ensure no errors are propagated. The pipeline is reproduced in full in **Appendix E**.

Analysis

All MRI-derived cross-sectional areas will be used as an alternative method for normalizing power specific power. For example, knee power in each task will be expressed as W/cm^2 where the cross-sectional area refers to that of the knee extensors. These results contribute to additional findings in Research Question #1.

The MRI DTI parameter FA was of principal interest due to its prevalence in muscle diffusion literature, particularly related to muscle power and fibre type. Linear regression was used to determine if ankle power derived via inverse dynamics during a CMJ, could be predicted by FA of the soleus, lateral gastrocnemius or medial gastrocnemius. Analysis via 95% CI using a t-distribution was used to determine if there were any significant differences in FA of the major ankle musculature (soleus, lateral gastrocnemius, medial gastrocnemius or tibialis anterior) between athletic disciplines (short-distance, long-distance and non-athletes). A final regression analysis attempted to uncover whether a significant linear relationship existed between muscle fiber proportion (taken from existing literature) and the average FA within that muscle.

Results

Predictive Ability of FA

A linear regression model investigating the predictive ability of soleus FA on ankle specific power (W/kg) during the CMJ generated a statistically significant predictive model ($F_{1,32}$ = 4.324, p=0.046, r = -0.350). FA for lateral gastrocnemius ($F_{1,34}$ = 0.011, p=0.916) and medial gastrocnemius ($F_{1,35}$=0.603 p=0.443; Hypothesis 5.1) failed to generate a statistically significant model. The relationship between ankle power and FA of the soleus is shown in **Figure 8.2**. Note that the dependent variable (ankle power) is shown on the abscissa to facilitate comparison to existing literature which has organized data similarly. Relatedly, the effect of age on diffusion-based MRI methods has been investigated with differing results (see **Table 8.4**). Although not driven by a specific hypothesis, considering the wide age-range in our cohort, it was prudent to investigate whether the effect of age confounded FA in our cohort, and thus an additional regression analysis was conducted. The effect of age on soleus FA was investigated and failed to generate a statistically significant model ($F_{1,32}$ = 0.405, p=0.529). Thus, there was no statistically significant linear relationship between age and soleus FA in this cohort.

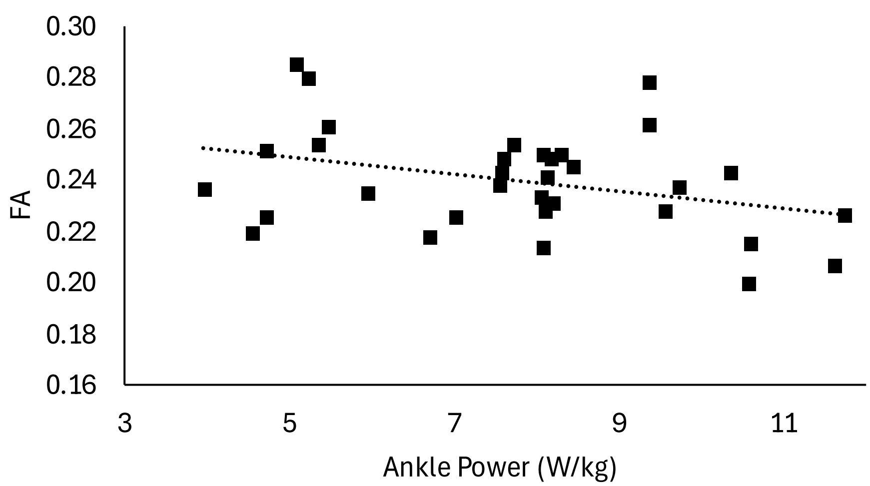

Figure 8.2 – The relationship between ankle power derived via inverse dynamics during a CMJ and MRI-derived fractional anisotropy (FA). FA of plantarflexors successfully predicted ankle power during a CMJ ($F_{1,32}$ = 4.324, p=0.046, r = -.350). This finding supports the linear relationship between plantarflexor FA and ankle power during a CMJ. FA of the soleus depicted here.

The Effect of Athletic Discipline

With respect to the effect of athletic discipline on FA, analysis via 95% CI (with t-distribution) showed that short distance athletes had significantly lower FA compared to non-athletes in the soleus (short 95% CI: 0.206 to 0.233 vs. non-athlete 95% CI: 0.234 to 0.255) and the medial gastrocnemius (short 95% CI: 0.202 to 0.220 vs. non-athlete 95% CI: 0.224 to 0.253) (**Figure 8.3**). The six remaining confidence intervals (i.e., short-distance tibialis anterior to non-athlete tibialis anterior etc.) demonstrated overlap. Thus, there appears to be a small effect of athletic discipline on shank muscle FA, particularly the plantarflexors soleus and medial gastrocnemius. Although not directly related to a hypothesis, tibialis anterior appeared to have consistently higher FA in all athletic disciplines, compared to other muscles.

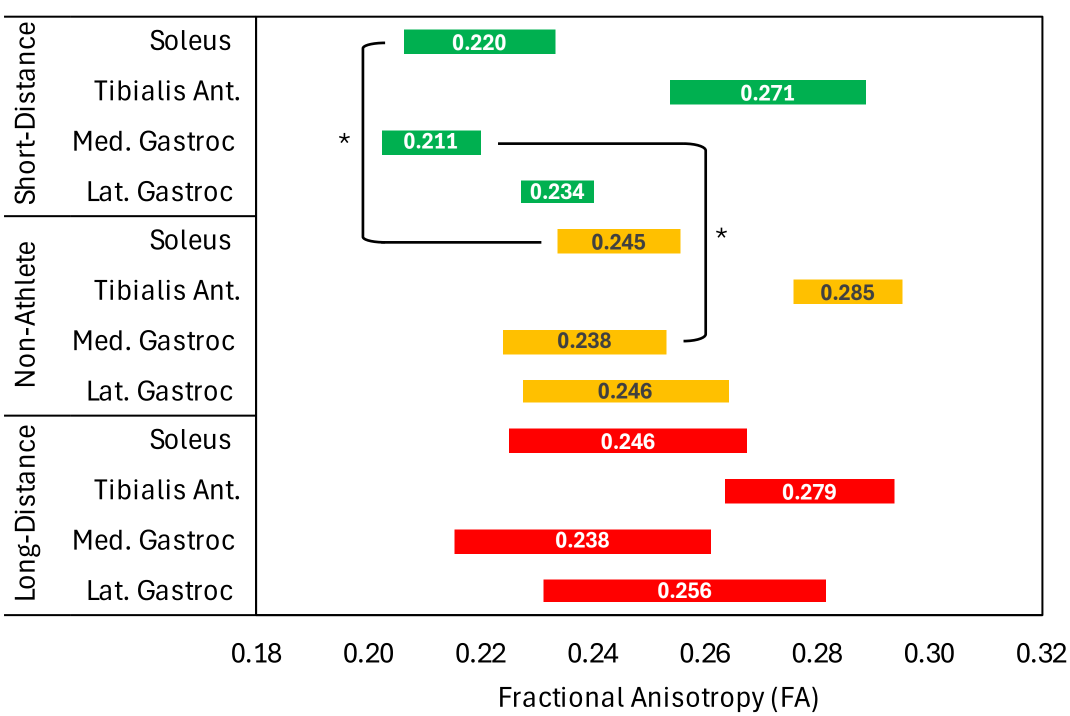

Figure 8.3 – 95% CIs for all shank muscles analyzed, organized by athletic discipline. The short distance CIs are shown in green, non-athletes CIs in orange and long-distance Cis in red. Short distance athletes had lower FA in the soleus and medial gastrocnemius compared to non-athletes. Of note, tibialis anterior FA differed significantly from all plantar flexor muscles in short-distance and non-athletes. 95% CIs are calculated with a t-distribution due to small sample sizes. Data labels represent the mean.

FA and Muscle Fiber Proportion

Muscle fiber composition from existing literature was predicted to show a negative correlation with fractional anisotropy (see **Table 8.5** for muscle fiber composition) as was observed for thigh muscle DTI in Cameron and colleagues (2023). The present regression analysis failed to generate a statistically significant predictive model for FA based on fiber type ($F_{1,5} = 0.343$, $p=0.589$; **Figure 8.4**).

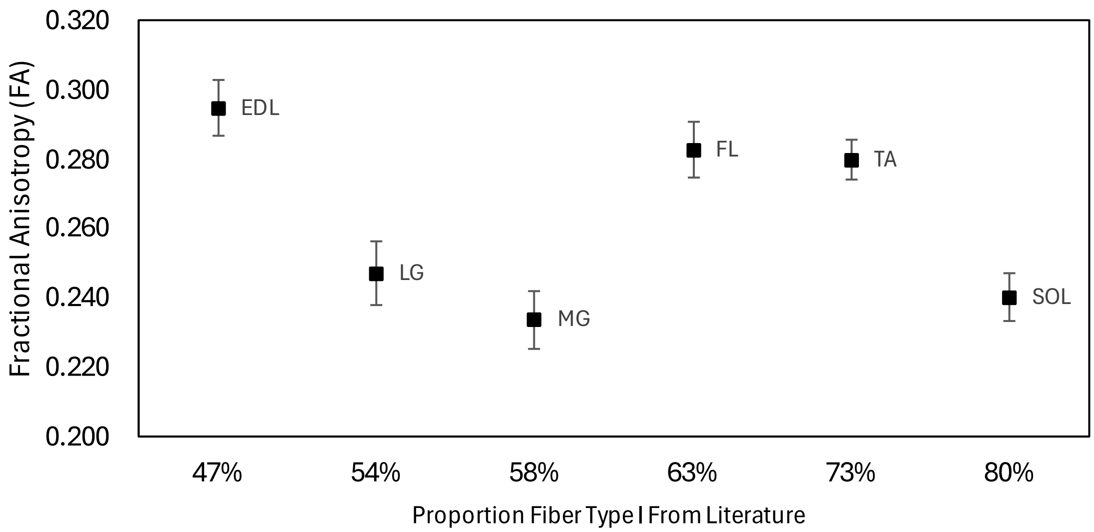

Figure 8.4 – Fractional anisotropy (FA) for muscles of the shank sorted by type I fiber proportion. Our regression model was not statistically significant, and therefore fiber type failed to predict FA in our cohort. Data points are mean values and error bars are 95% CIs. Fiber type proportions are obtained from existing literature (see **Table 8.5**). EDL: extensor digitorum longus, LG: lateral gastrocnemius, MG: medial gastrocnemius, FL: fibularis longus, TA: tibialis anterior, SOL: soleus.

Key Findings

- FA of the soleus successfully predicted ankle power derived via inverse dynamics, measured during a CMJ. This agrees with the findings of Scheel and colleagues (2013) where ankle power was measured via Biodex.
- Short distance athletes showed lower FA in soleus and medial gastrocnemius.
- Non-significant regression model relating fiber type from literature and measured FA. This contrasts with significant negative correlation found by Cameron and colleagues (2023).

Discussion

Overview

This section presents the fractional anisotropy results in the context of existing data related to muscle power and fibre type. The issue of sample size is discussed, as are confounding factors such as age, sex and ankle position. These confounding factors likely play a part in each section.

Predictive Ability of FA

In 2013, Scheel and colleagues investigated the relationship between several quantitative MRI parameters and muscle fiber composition in 12 males aged approximately 26 years. They performed MRI-DTI imaging in the shank followed by fine-needle biopsy and histology to determine the number and size of type I and II muscle fibers. Fractional anisotropy (FA), which describes the degree to which diffusion occurs in a single direction (with a value of 1 being highly aligned diffusion), was greater in those with higher Type I fiber proportion (R^2 =0.5, p=0.01). The authors suggested that Type I fibers produced a larger diffusion hindrance in the radial direction, possibly due to cellular microstructures. This would result in greater diffusion in the axial direction, and thus, greater FA. A similar study by the same group (Scheel et al., 2013a) conducted MRI DTI imaging and evaluated soleus-derived ankle power using a Biodex in 11 males, aged approximately 27 years. The knee was bent to approximately 40 degrees to minimize the contribution of other plantarflexors. They found that FA was strongly negatively correlated with ankle power output (-0.85, p=0.0015). In accordance with their previous work, the authors reasoned that the lower FA and greater power implied greater Type II fiber composition. As well, they suggested that the larger fiber diameter and lower mitochondrial density in a Type II fiber may explain the increased radial diffusivity observed, and therefore the lower FA as well (as FA is calculated with RD – See **Appendix F** for specific formulae). Takao and colleagues (2022) assessed MRI-DTI of the soleus and tibialis anterior in ten males, aged approximately 23 years using three contraction conditions: during a neutral rested state, during neutral isometric contraction and during a shortened isometric contraction. They measured ankle power with a Biodex (knee angle was not described) and assessed the effect of passive muscle shortening and lengthening of the tibialis anterior during plantarflexion on DTI parameters. Takao and colleagues found no statistically significant correlation between soleus FA and ankle power in any of the ankle positions or contraction conditions (Takao et al., 2022, $p>0.305$), including a neutral position similar to the ankle position used in this study.

The results from this study are therefore consistent with that of Scheel and colleagues, but contrary to Takao and colleagues. This study demonstrated a statistically significant linear relationship between FA of the soleus and ankle power during a CMJ ($r=-0.350$) which was not as strong as that observed by Scheel and colleagues ($r=-0.85$) but consistent in direction. The strength of their relationship may result from the small sample size which can increase the likelihood of false positives or negatives, especially in the presence of high variance (Jenkins and Quintana-Ascencio, 2020). As well, the age of the cohort in the present study varied greatly. Despite no discernible relationship between age and soleus FA in this study ($p=0.529$), other studies have observed such a relationship. Galbán and colleagues in 2007 investigated the effect of age on diffusion parameters in 38 males aged approximately 27 to 67 years. They found a significantly greater tibialis anterior FA ($p<0.05$) in young adults, but no significant age effect in the plantarflexors. Cameron and colleagues (2023) found the opposite age effect – they found that whole-thigh FA increased with age ($\beta = 0.33$, $p=0.001$). Sex is also known to affect FA – Males have been shown to have greater FA values in the plantarflexors than females by 2-13% (Galbán et al., 2005). Our cohort consisted of 23 males and 13 females; therefore, the average FA could be expected to be lower than that of Scheel and colleagues, who had a male-only sample. The effect of joint position, which drives passive and active tissue elongation, has been shown to affect FA. Schwenzer and colleagues (2009) detected increased tibialis anterior

FA when placed in a plantarflexed position (passively lengthened) and increased plantarflexor FA when placed in a dorsiflexed position (passively lengthened). They authors concluded that stretching of the muscle caused a narrowing of the fiber radius thereby inhibiting diffusion in the radial direction, increasing FA. Our participants did not have their ankle joints constrained as was done by Scheel and colleagues, which may explain some of the differences observed in our results. In summary, the effect of sample size, age, sex, ankle position likely contributes to the differences between our findings and that of existing literature.

Curiously, no other plantarflexors demonstrated a relationship between FA and ankle power. This may be due to differences in the contribution to CMJ performance from various shank muscles. Previous simulation studies have demonstrated greater (modeled) muscle tension (Nagano et al., 2007), greater contribution to vertical velocity (Suzuki et al., 2017) and greater muscle activation via surface EMG (Bb and Dsouza, 2024) in the soleus versus gastrocnemii during a maximal CMJ. These previous findings demonstrate that the soleus plays a large role in CMJ performance and thus may explain why only soleus FA was able to predict ankle power output in this study.

The Effect of Athletic Discipline

On the basis that FA was able to predict muscle fiber type composition (Scheel et al., 2013b) and power output (Scheel et al., 2013a), it was expected that FA would also be significantly lower in short distance athletes. Our results showed that medial gastrocnemius and soleus FA in short distance athletes was significantly lower than that of non-athletes indicating a lower proportion of Type I fibers in short distance athletes, however 6 other confidence intervals comparisons failed to indicate a significant difference between athletes.

A possible confounding factor in this comparison is that of age. The short-distance athletes were on average 65 years of age compared to 59 years of age for long-distance athletes and 56 years of age for non-athletes. No significant age effect was noted in this cohort, as mentioned above, however Galbán and colleagues (2007) observed lower FA in the tibialis anterior of older adults along with no significant age-effect in the plantarflexors. The extent of the age-effect on diffusion parameters appears to be small but also inconsistent so its effects as a confounder are likely minimal.

A second confounding factor in this investigation may be that of sex as the short distance athlete cohort had only one female. In 2005, Galbán and colleagues determined that males showed 5-13% greater FA in the lateral gastrocnemius, and minimal differences in the tibialis anterior (2%). Interestingly, the effect observed in **Figure 8.3** shows the opposite effect, with the plantarflexors of the short distance athletes showing lower FA than other athletic disciplines (whereas a higher proportion of males should have shown increased FA according to the work of Galbán and colleagues in 2005). It appears most likely that other currently unidentified factors are confounding these results.

FA and Muscle Fiber Proportion

A creative investigation by Cameron and colleagues in 2023 organized literature-based measures of fiber type proportion in the thigh and compared their measured FA values. Interestingly, they found decreasing FA with increasing type I proportion (β=-0.7, p= 0.024). This contrasts with the findings of Scheel and colleagues (2013) who found greater FA with greater proportion of Type I muscle fibers via needle biopsy. Our results showed decreasing FA with increasing Type I fiber proportion which agrees with Cameron and colleagues (2023); however, the model was not statistically significant. The opposite findings presented by Scheel and colleagues, versus that of Cameron and colleagues are intriguing. The work of Scheel and colleagues represents the only direct assessment of MRI DTI and muscle biopsy together and is likely methodologically stronger however it stands to be improved by a larger sample size with a greater age range.

A potential explanation for the unsuccessful regression model relates to ankle position. Previous research has demonstrated that passive ankle plantarflexion results in greater FA of the tibialis anterior. **Figure 8.4** shows that FA of the tibialis anterior in this research was high despite being composed of approximately 73% type I muscle fibers according to existing literature. This research employed a natural ankle position and thus the FA measured in the tibialis anterior may be artificially inflated by some degree of passive plantarflexion, contributing to regression model error.

Table 8.4 – Summary of existing literature on the topic of fractional anisotropy (FA) in muscle, and the various factors that can affect its measurement.

Reference	Sample	Age	Findings	Notes
Scheel et al., 2013a	11m	27.2 ± 10.7	-Correlation of soleus-isolated ankle power and FA (r=-0.85, p=0.0015), and RD (r=0.80, p=0.0047).	-Isokinetic pflex with sharply bent knee, 90° ankle, 7 different ω -1.5T MRI, 90° ankle position
Scheel et al., 2013b	12m	26.0 ± 10.1	-Correlation of type 1 fiber proportion and FA (r=0.707, p=0.010), MD (r=-0.612, p=0.035) and RD (r=-.721, p=0.008).	-Histochemical staining -1.5T MRI, 90° ankle position
Takao et al., 2022	10m	21-27	-Correlation between soleus FA and maximum work in non-contracted state (r=0.67, p=0.003). -No correlation between soleus FA and power in non-contracted (r=0.04, p=0.907), isometric (r=0.11, p=0.753) or shortened state (r=0.36, p=0.305).	-Isokinetic pflex and dflex @ 30, 60, 120°/s -3T MRI with various contractions Non-contracted, neutral ankle Isometric, neutral ankle Concentric, max plantarflexion
Galbán et al., 2004	6m	--	-Correlation between λ_3 (R^2 = 0.74) and FA to PCSA (R^2 = 0.89). -Negligeable effect of fiber length on diffusion	-1.5T MRI -PCSA of shank from MRI, architecture from literature
Galbán et al., 2005	12m, 12f	21-39	-FA in pflex higher in males (2-13%)	-1.5T MRI
Galbán et al., 2007	38m	46.4 ± 14.1	-All eigenvalues for pflex muscles lower in old participants -No sig diff in FA in old vs young plfex -FA lower in old adult dflex	-1.5T MRI -Old vs young
Esposito et al., 2013	14f (mice)	2-17.8mo	-FA greater in old mouse TA and gasctroc -FA decreases in response to injury and subsequently rebounds.	-7T MRI -Old vs young, plus injury induced -Histological examination
Cameron et al., 2023	51m, 43f	22-89	-FA whole-thigh increases with age (β = 0.33, p=0.001) -MD whole-thigh decreases with age (β = -0.36, p<0.001) -FA and MD diff b/w thigh muscles - Decreasing FA with increasing type I proportion (β=-0.7, p= 0.024; Supplemental S1)	-3T MRI -Knee flexor and extensor torque -Old vs young -Compared to existing cadaveric fiber type data
Hatakenaka et al., 2008	5m	23-36	-FA of tib ant and med gastric are diff in passive pflex vs dflex	-1.5T MRI -Passive pflex and dflex
Schwenzer et al., 2009	5m, 3f	29 ± 7	-FA of tib ant greater in pflex vs dflex -FA of sol greater in dflex vs pflex -FA of med and lat gastric greater in dflex vs pflex	-3T MRI -Pflex, neutral and dflex
Okamoto et al., 2010	4m, 6f	25-51	-FA of gastroc increased with active pflex -FA of tib ant no change with active pflex	-1.5T MRI
This Research	23m, 13f	57.8 ± 17.6	FA no relationship with age Relationship with athlete type Relationship between muscles	Neutral, unconstrained ankle

Table 8.5 – Summary of the existing literature used to formulate average type I muscle fiber proportions. † indicates average of superficial and deep samples.

Reference	Sample	Age	EDL	Lat. Gastroc.	Med. Gastroc.	Fib. Long.	Tib. Ant.	Soleus
Edgerton et al., 1975	22m, 10f	59 ± 3		50 ± 4%	52 ± 3%		--	70 ± 4%
Gollnick et al., 1974	9m, 2f	24-41		60.2% (45-82)	--		--	80.4% (64-100)
Johnson et al., 1973 †	6m	17-30	47.3 ± 5.2%	46.9 ± 6.9%	50.8 ± 4.2%	62.5 ± 9.6%	73.0 ± 7.8%	87.7 ± 9.6%
Elder et al., 1982	4m	20-27		--	--		--	76% (60-92)
Green et al., 1981	10m	20-24		49.4 ± 2.8%	--		--	--
Henriksson-Larson et al., 1983	5m	18-32		--	--		72%	--
Dahmane et al., 2005 †	15m	17-40		--	69.7%†		73.0%†	88.0%†
Average			**47.3%**	**54.1%**	**57.5%**	**62.5%**	**72.7%**	**80.4%**

Revisiting Hypotheses

5.1 Fractional anisotropy (FA) of the soleus, lateral gastrocnemius and medial gastrocnemius muscle would be inversely proportional to ankle power output.

Partially accepted.

5.2 Short distance athletes will show lower FA in the plantarflexors compared to both non-athletes and long-distance athletes.

Partially accepted.

5.3 There will be a linearly decreasing correlation between the proportion of type I muscle fiber in each muscle and the measured FA for that muscle.

Rejected.

Conclusion to RQ #5

Fractional anisotropy (FA) of the soleus weakly predicted ankle power during the CMJ. This finding agrees with foundational literature on the topic. The lower strength of the linear correlation presented here may result from the larger sample size and age-range of the cohort. Differences in FA of the medial gastrocnemius and soleus in short distance athletes versus non-athletes was uncovered. Based on previous literature relating FA to Type I muscle fiber proportion, this finding suggests that short distance athletes have a lower proportion of Type I fibers in the medial gastrocnemius and soleus compared to non-athletes, which is conceivable. It is curious however that non of the other pairwise comparisons uncovered a similar pattern, but this may be due to different mean age among the athletics discipline, or a lack of statistical power in the short and long-distance athlete groups. The comparison of FA to existing literature fiber type proportion has previously been successful however this research failed to generate a successful model potentially due to the natural ankle position used in this research. Overall, the application of diffusion imaging to approximate muscle power capacity or muscle fiber type appears feasible but challenged by confounding factors not limited to age, sex and joint position.

9 Overall Discussion and Conclusions

Age, sex and PA measured by the HUNT questionnaire were predictive of lower-body power in the CMJ. Remarkably, the magnitude of effect for activity level was comparable to that of age itself. Activity level had a positive protective effect on lower-body power in addition to knee and hip joint power during the CMJ however ankle power during the CMJ was not successfully predicted by activity level. This may be due to the large number of masters athletes in our cohort and a presumed training focus on those larger muscle groups. The power “loss” through the lifespan was comparable to existing literature (approx. 50%) but the annual “loss” rates dependent on the reference age, type of jump and the technology used to measure power.

Athletic discipline had a positive protective effect on lower-body power in addition to ankle, knee and hip joint power during the CMJ with short-distance athletes having the highest predicted power. In contrast, short-distance masters athletes also had the steepest rate of power “loss” compared to long distance and non-athletes. This implies that short distance athletes are predicted to have higher power output at a given age but are also expected to lose power at a greater rate. Power outputs between short distance athletes and all others can be expected to coincide eventually but this is not projected to occur in this model within a normal lifespan. This finding agrees with existing literature showing approximate power “loss” of 50-60% over the course of 50 years in both long- and short-distance masters athletes, despite the steeper “loss” rate.

No effect for age, sex or activity level was found on changes in power from the pre to post condition in the functional task circuit. This finding contrasts with existing literature on the topic of fatigue in older adults, although much of this literature was conducted with single joint power assessments. Other existing literature related to fatiguing protocols suggests that although the circuit was representative of real life, it was likely insufficiently challenging as a result.

Trunk angle, measured at the instant of peak GRF, was found to predict lower-body specific power. While this finding has been demonstrated in the literature, a novel addition is that predominantly younger adults were found to use this strategy, whereas older adults did not. Thus, in addition to the known, physiologically driven implications of age on power output, it appears that movement strategies can have a measurable effect as well, particularly when using a complex movement such as the CMJ,

Lastly diffusion MRI in the soleus muscle successfully predicted ankle power, although the relationship was not as strong as previously observed in existing literature. As well, differences in FA of the medial gastrocnemius and soleus were found in short distance athletes versus non-athletes. Based on previous literature relating FA to Type I muscle fiber proportion, this finding suggests that short distance athletes may have a lower proportion of Type I fibers in the soleus and medial gastrocnemius compared to non-athletes. Different mean age among the athletic disciplines, which is known to affect diffusion parameters may help explain why other pairwise comparisons did not uncover a similar pattern. While more research is needed to understand confounding variables, these findings suggest that diffusion MRI may be capable of non-invasively linking measures of cellular diffusion, and possibly cellular structure, to function.

Overall, while age consistently demonstrated negative effects on muscle power, activity level and athletic discipline showed positive, protective effects on lower-body power, and on joint power during the CMJ. The magnitude of effect for activity level was comparable in magnitude to the effect of age on lower-body power output. Trunk posture was also shown to affect lower-body power output, with further analysis showing that this strategy was mostly adopted by younger adults. Therefore, age-related reductions in power during the CMJ may be due in part to technical considerations. Throughout the functional task circuit, no decrease in power was observed. It appears most likely that the circuit was insufficiently challenging for our cohort. The strong positive effect of PA level on muscle power output in older adults, which was especially strong in short-distance athletes suggests that short-distance athletics may be an interesting and enjoyable way for older adults to maintain power output.

Future Directions

Future research should determine if the greater lower-body joint power demonstrated by short-distance athletes in the CMJ shows any benefit in terms of reducing the burden of functional tasks. As well, future investigations into muscular fatigue of older adults should consider the use of single joint power measurements in addition to multi-joint movements such as the CMJ. The application of a more challenging CMJ protocol such as the 30s Bosco test may yield more clear fatigue effects, however, this may come at the expense of contextual validity.

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Appendix A: Kinematic and Kinetic Summary of Functional Tasks

Table A1 – Sagittal plane kinematic and kinetic data during stair ascent.

Reference	Sample	ROM (°)			Peak Specific Net Torque (N-m/kg)			Peak Power (W/kg)			Notes
		Ankle	Knee	Hip	Ankle (+pflx)	Knee (+ext)	Hip (+ext)	Ankle	Knee	Hip	
Hortobagyi et al., 2003	n=13, mean 22 n=14, mean 74	---	57 ± 6 53 ± 3	---	---	1.55 ± 0.24 1.00 ± 0.22	---	---	---	---	Knee angular velocity at peak torque higher in young vs old (122 ± 17 vs 105 ± 19°/s) 19cm riser, 27cm tread
Nadeau et al., 2003	n=11, 41-70	39.2	83.1	55.4	1.17 ± 0.14	0.98 ± 0.18	0.53 ± .17	2.53 ± 0.52	1.79 ± 0.50	1.01 ± 0.48	33 deg slope – 17cm riser, 26cm tread
Kirkwood et al., 1999	n=30, mean 65.4	---	---	---	---	---	1.00	---	---	---	Peak rate of change in moment 0.090°/s
Karamanidis and Arampatzis, 2009	n=16, mean 28 n=28, mean 64	---	---	---	1.7 1.4	1.5 1.2	0.5 0.7	---	---	---	Graphs appear to have at least one error
McFadyen and Winter, 1988	n=3, mean 83	~32	~90	~50	~1.6	~1.4	~1	~4.55	~3.34	~1.98	All values interpreted from graphs (~)

Table A2 – Sagittal plane kinematic and kinetic data during stair descent.

References	Sample	ROM (°)			Peak Specific Net Torque (N-m/kg)			Peak Power (W/kg)			Notes
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip	
Hortobagyi et al., 2003	n=13, mean 22 n=14, mean 74	---	82 ± 2 82 ± 6	---	---	0.90 ± 0.24 0.64 ± 0.24	---	---	---	---	Knee angular velocity at peak torque higher in older vs younger (-56 ± 26 vs -25 ± 26°/s)
Kirkwood et al., 1999	n=30, mean 65.4	---	---	---	---	---	0.50	---	---	---	Peak rate of change in moment 0.081°/s
McFadyen and Winter, 1988	n=3, mean 83	~43	~85	~23	~1.3	~1.5	~0.3	~6.69	~6.38	~0.76	All values interpreted from graphs (~)

Table A3 – Sagittal plane kinematic and kinetic data during chair rise.

References	Sample	ROM (°)			Peak Specific Net Torque (N-m/kg)			Peak Power (W/kg)			Notes
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip	
Hortobagyi et al., 2003	n=13, mean 22 n=14, mean 74	---	81 ± 4 82 ± 8	---	---	0.84 ± 0.18 0.69 ± 0.13	---	---	---	---	Knee angular velocity at peak torque higher in young vs old (51±14 vs 41±12°/s)
Roebroeck et al., 1994	n=10, mean 27	~15	~93	~100	~0.30	~0.48	~0.35	---	---	---	Peak moments presented as N-m/kg-m
Carr and Gentile, 1994	n=6, mean 24	~18	~80	~90	1.11	1.19	3.05	~0.3	~1.6	~1.2	Some values interpreted from graphs (~)
Rodosky et al., 1989	n=10, mean 25.5	~18	~85	~70	---	---	---	---	---	---	Some values interpreted from graphs (~)
Greve et al., 2013	n=14, mean 24.3 n=10, mean 76.4	---	---	---	---	2.24 ± 0.29 2.02 ± 0.25	2.4 ± 0.36 2.21 ± 0.36	---	---	---	

Appendix B: Participation Screens

Informed Consent Form

Informed Consent Form (ICF)

Date: ____________________

Study Name: *The Power of Exercise: the effect of age and activity level on muscular power*

Researchers:
Lead Researcher: Dan Desroches, PhD Candidate in Kinesiology & Health Science
2020 Sherman Health Science Research Centre
desroches.dan@gmail.com

Principal Investigator: Dr. Janessa Drake, Associate Professor
2030 Sherman Health Science Research Centre
416-736-2100 Ext. 33568
jdrake@yorku.ca

Purpose of the Research: This research will assess the effect of various activity levels on maximal power output, and power output during various everyday tasks. This information will hopefully help us to improve our understanding of the types and quantities of activities that are important to preserve function as we age.

What You May Be Asked to Do in the Research: This study consists of two lab sessions plus an optional additional two sessions in 6-month time.

- *In the first session*, you may be asked to complete a series of range of motion tasks and some endurance assessment tasks such as forward bending and a plank. You may be asked to complete a vertical jump assessment to measure your maximal lower body power output. This task involves a warm-up, familiarization to the task, and several maximum-effort attempts with adequate rest. Following the vertical jump assessment, you may complete activity level surveys. Following the surveys, you may perform several functional tasks. Examples of a functional task would be rising from a chair five times in a row, or a 3-to-5 stair climb and descent, 6min walk, box lifts from floor to waist height (10% body weight), side lying leg raise, etc. Motion capture and force sensing equipment will provide detailed insight into how these tasks are accomplished by you and other individuals of varying activity levels. We will ask that you wear a heart rate monitor strap and a breath analyzer (VO2 mask) during some activities. This session will be ~2hr in duration.
- *In the second session*, if you qualify for scanning you will undergo magnetic resonance imaging scans (MRI) of the lower limbs, pelvis, and lower spine region to assess musculoskeletal properties such as muscle quality, approximate muscle fiber type, and bony geometry in these regions. Following the MRI scans, you will complete a VO2 max assessment to assess your cardiorespiratory fitness. This session will be ~2hr in duration.
- *Sessions 3 and 4*, will be held after 6-month. Participants may choose to return and redo these same activities to assess the changes after a 6-month time span (we will ask that you complete another copy of this informed consent form at that time).
- *Throughout the time between sessions 2 and 3*, we ask you to wear an activity monitor for seven consecutive days of typical activity (1 week). Participants can also choose to wear the monitor for two days/month between sessions 2 and 3. We will provide you with the activity monitor device (Actigraph GT9X) and we will download the stored data from the devices when you return it to us. Your name and location are not recorded on the devices. The activity monitor can be returned before or at session 3. Session 3 and 4, including the activity monitoring are optional. The total time commitment in the lab is approximately 4-8 hours (depending on sessions that are completed), not including travel time. If you would like to share your Garmin, Apple, Fitbit, or Strava data with us, you are welcome to do so, but are not required to do so to participate in this research. It is up to you if you would like to have a check-in and/or reminder from the researcher regarding the use of the activity monitor devices.

  Inducements: After we determine you are eligible to participate, you will be given $20 for each of the 4 sessions you start, as well as $10 for transit or a free parking pass for each campus visit. Please note, if you are not able to complete session 1 and/or 2, we may not schedule sessions 2, 3, or 4, and/or may not ask you to complete the activity tracking tasks.

Risks and Discomforts: We anticipate minimal real or perceived risk to you from participating in this study.

During the collection, you will be asked to exert yourself at a high level which can pose a physical risk to you. To mitigate these potential risks, *all participants are advised to consult their physicians prior to participating in our research.* You can choose to be connected to our fall-restraint system during any/all standing based tasks. You will be given ample time to complete a guided warmup before beginning any strenuous physical activity. You will also be given generous rest periods, and will be strictly attended to by research assistants in case of light-headedness, loss of balance, etc. There will always be at least two researchers in the lab so that help is available in the event of an emergency. You may experience soreness the day following the assessments, likely lasting 1-2 days after the study as would be expected with physical activity, and we ask that you contact your healthcare provider should their discomfort persist longer and to inform us. Although uncommon, it is possible for certain people to develop a temporary rash on your skin in response to the tape used to affix the markers to your skin (similar to a response to adhesive bandages). This is minor and resolves without treatment in a few days.

You may feel self conscious or uncomfortable during the physical exertion in our lab. We will guide you though a warmup and be informed of what you can expect to feel during the tasks. We can ask questions or choose not to complete any task. Your participation in the study is completely voluntary and you may choose to stop participating at any time.

We do not need to collect your name in this study, so we label all data with a randomly generated alphanumeric code. We will aggregate the data by 10-yr age groups and sex, and stratify by activity level so there is a minimal chance of you being identifiable in our research outputs. But we can't control the minimum number of participants in your group. As such, we have provided an opportunity for you to choose how you would like your data used with a series of checkboxes below (under Additional Consent Requests).

MRI Information and Related Potential Risks:

What is involved in this portion of the study?

Your participation will involve measuring the anatomy and activity of your spine, pelvis, and lower limb using MRI. MRI scanners image your body areas using radio waves and very strong magnetic fields. You will be asked to fill out a safety screening form to assess whether it is safe for you to enter the MR room. It is important that you provide us with an accurate and up-to-date medical history, and when unsure to ask clarifying questions so that we can proceed safely. You will then be asked to remove any metallic objects you may be carrying (for example, wallets, watches, earrings, or piercings) and possibly to change clothing into a gown that we will provide (if deemed necessary because of large zippers etc.). You will be required to lie completely still on the patient bed that will slide into the bore of the MRI scanner. You will be able to communicate with us at all times via a built-in intercom. You will be holding an emergency bulb that you can squeeze at any time to let us know you want to come out of the MRI scanner.

This is not a clinical evaluation.

The images of your spine, pelvis, and lower limbs collected in this study are not intended to reveal any disease state, in part because this MRI protocol is not designed for clinical diagnosis. Thus, your images will not be routinely examined by a clinical radiologist. The personnel at the MRI Research Facility are not qualified to medically evaluate your images. However, if in the course of collecting images we have any concerns, we may show your scans to a clinical radiologist, who may suggest that you obtain further diagnostic tests.

At the investigator's discretion, you may view your images and receive digital copies of them. However, you should be aware that anatomical structures within the normal population are highly variable, and that it is difficult to draw any conclusions from your images; you should be aware of the potential distress or discomfort that may occur by viewing your own images. Do not rely on this research MRI to detect or screen for abnormalities.

What are the risks of being scanned?

Metal: The MRI scanner produces a constant strong magnetic field, which may cause any metal implants and/or clips within their body to shift position. The magnetic field may also cause any implanted medical devices to malfunction. Thus, if you have any implanted metal, clips, or devices, it is hazardous to your health to participate in this study. Please provide us with as much information as you can, for example if you had surgery in the past, so that we may decide whether it is safe for you to be a subject. Metallic objects brought into the MRI environment can become hazardous projectiles. Metal earrings, body piercings, and necklaces must be removed prior to the study.

Pregnancy: Exposure to MRI scanning might be harmful to a pregnant female or an unborn child. Although there are no established guidelines at this time about MR and pregnancy, you should be informed that there is a possibility of a yet undiscovered pregnancy related risk. If you know or suspect you may be pregnant or if you do not want to expose yourself to this risk, we recommend that you do not participate in this study.

Inner ear damage: MRI scanning produces loud noises that can cause damage to the inner ear if appropriate sound protection is not used. Earplugs and/or headphones will be provided to protect your ears.

Claustrophobia: When you are inside the MRI scanner, the MRI scanner surrounds your body and your close-fitting scanning coils may be positioned around or near your spine, pelvis, and lower limbs. If you feel anxious in confined spaces, you may not want to participate. If you decide to participate and begin to feel claustrophobic later, you will be able to tell us via the intercom and we will discontinue the study immediately.

Burns: In rare cases, contact with the MRI transmitting and receiving coil, conductive materials such as wires or other metallic objects, or skin-to-skin contact that forms conductive loops may result in excessive heating and burns during the experiment. The operators of the MRI scanner will take steps, such as using foam pads when necessary, to minimize this risk. Tattoos with metallic inks can also potentially cause burns. Any heating or burning sensations during a scan in progress should be reported to the operators immediately and we will discontinue the scan.

Besides the risks listed above, there are no other known risks from the magnetic field or radio waves at this time. Although functional MRI scanning has been used for more than 15 years, long-term effects are unknown. If new findings about the risks of the MRI technique become available within a year of your participation, we will let you know about them.

Benefits of the Research and Benefits to You: The benefits of the research are numerous. Age has been implicated as a major cause of performance decline in humans. Recent research suggests that age may accompany performance declines, but it does not cause them, per se. Instead, it is suspected that inactivity (which is known to be an issue in older adults) is a primary driver of performance decline across the lifespan.

One major performance variable that has received much attention lately is muscular power. Research has demonstrated that older adults with low muscular power consistently underperform on functional task assessments. At the same time, older adults who train to improve muscular power can improve functional task performance scores. This relationship likely holds for younger adults. Muscular power appears to be a critical aspect of healthy aging.

Your participation is instrumental in gathering this knowledge. Additional benefits to the participant would be access to your MR images and a maximum lower body power assessment. You will also get to experience laboratory grade motion capture software.

Voluntary Participation and Withdrawal: Your participation in the study is completely voluntary and you may choose to stop participating at any time. Your decision not to volunteer, to stop participating, or to refuse to answer particular questions will not influence the nature of your relationship with the researchers and York University either now, or in the future.

If you stop participating, you will still be eligible to receive the $20 for each session you attempt, even if you withdraw without completion of the session. In the event you withdraw from the study, all associated data collected will be immediately destroyed wherever possible. You will also be provided with a $10 transit fee or free parking pass for each session you attempt. Should you wish to withdraw after the study, you may do so up until the analysis is complete.

Confidentiality: All information you supply during the research will be held in confidence and unless you specifically indicate your consent, your name will not appear in any report or publication of the research. Unless you indicate your consent otherwise below, your data will only be reported in an aggregated manner (by age decade, sex, activity level) in a group not smaller than 3 participants. In no case will your personal information be shared with any other individuals or groups without your expressed written consent. If you choose to provide opinions and/or anecdotes to the researcher, you may choose to indicate your consent below for their inclusion in the dissemination of the research (should it be applicable).

The principal investigator will keep a link that identifies you to your coded information, but this link will be kept secure and available only to the principal investigator and/or selected members of the research team. Any information that can identify you will remain confidential.

Data will be stored on a hard drive on a password-protected computer inside a locked laboratory, in a secure research facility. During processing and analysis, some processed data will be located on a personal laptop, secured by password. Any hard copies of coded data (e.g. questionnaires) will be stored in the cabinet in the same locked laboratory. Your MR images will be stored on secured computer servers and will be archived indefinitely. Consent forms will be stored in a locked cabinet in the same lab or in Dr. Drake's office. Only research staff/research team members will have access to this information. In all circumstances, confidentiality will be provided to the fullest extent possible by law.

The data collected in this research project may be used in anonymized form by members of the research team in subsequent research investigations exploring similar lines of inquiry. Such projects will still undergo ethics review by the HPRC, our institutional REB. The experimental data acquired in this study may in an anonymized form that cannot be connected to you, be used for teaching purposes, be presented at meetings, published, shared with other scientific researchers or used in future studies. Your name or other identifying information will not be used in any publication or teaching material without your specific permission. Any secondary use of anonymized data by the research team will be treated with the same degree of confidentiality and anonymity as in the original research project.

Please note that at the end of the study, anonymized may be deposited into one or more publicly accessible scientific repositories, such as York University Dataverse, an institutional research data repository, managed by York University Libraries and provided by Scholars Portal on behalf of the Ontario Council of University Libraries (OCUL), through which researchers from around the world will have access to these data for future research, through a [CC, CC-BY, CC-BY-NC, or other] standard data sharing license.

York University Dataverse does NOT accept content that contains confidential or sensitive information. Dataverse can be used to share de-identified and non-confidential data only. Contributors are required to remove, replace, or redact such information from datasets prior to upload. Scholars Portal makes backup copies of the uploaded data regularly in the event of a server or system malfunction, malicious attack, or other technical issues.

Questions About the Research? If you have questions about the research in general or about your role in the study, please feel free to contact Dan at 905-399-4671 or desroches.dan@gmail.com. This research has received ethics review and approval by the Human Participants Review Sub-Committee, York University's Ethics Review Board and conforms to the standards of the Canadian Tri-Council Research Ethics guidelines. If you have any questions about this process, or about your rights as a participant in the study, please contact the Director, Research Ethics in the Office of Research Ethics, 3rd Floor, Kaneff Tower, York University (e-mail ore@yorku.ca).

Legal Rights and Signatures:

I______________________, consent to participate in '*The Power of Exercise*' conducted by *Dan Desroches.* I have understood the nature of this project and wish to participate. I am not waiving any of my legal rights by signing this form. My signature below indicates my consent to participate.

Signature ____________________________ **Date** ______________________
Participant

Signature ____________________________ **Date** ______________________
Principal Investigator

Additional consent requests:

1. Video recording

I, ____________________ consent to the use of images of me (including video and other moving images), my environment and property in the following ways (please check all that apply):

In academic articles	☐ N	☐ Y	☐ Only if my identity is obscured
In print, digital and slide form	☐ N	☐ Y	☐ Only if my identity is obscured
In academic presentations	☐ N	☐ Y	☐ Only if my identity is obscured
In media	☐ N	☐ Y	☐ Only if my identity is obscured
In thesis materials	☐ N	☐ Y	☐ Only if my identity is obscured

2. Photographs

I, ____________________ consent to the use of images of me, my environment and property in the following ways (please check all that apply):

In academic articles	☐ N	☐ Y	☐ Only if my identity is obscured
In print, digital and slide form	☐ N	☐ Y	☐ Only if my identity is obscured
In academic presentations	☐ N	☐ Y	☐ Only if my identity is obscured
In media	☐ N	☐ Y	☐ Only if my identity is obscured
In thesis materials	☐ N	☐ Y	☐ Only if my identity is obscured

3. Data inclusion

The data we collect is labelled with a randomly generated alphanumerical code. We do not need your data identified by your name for this study. We will aggregate the outcomes by age/activity level, but we can't control the minimum number of participants in your group. Although remote, having a group of 2 or less participants may increase the potential risk of you being identified from the de-identified data. Likewise, we are providing you the opportunity to choose how your data is included in the resulting research publications.

I, ____________________________ consent to the use of my data in this study (please check all that apply):

Only if there are 3 or more participants in my group	☐ N	☐ Y
If there are 2 or less participants in my group	☐ N	☐ Y
Even if I am the only participant in my group	☐ N	☐ Y

4. Consent to waive anonymity

I, ____________________________ consent to the use of my name in the publications arising from this research.

5. Consent to data deposit

I understand that my de-identified data will be placed into an open research data repository. Y / N

6. Consent to use of quotes

I, ____________________________ consent to the use of quotations in all dissemination of the research (please check all that apply):

If they are de-identified	☐ N	☐ Y
If I am named	☐ N	☐ Y

7. Consent to be contacted regarding other research opportunities

I understand that the researcher of this study may contact me regarding other research studies. Y / N

CSEP approved Sept 12 2011 version

PAR-Q+

The Physical Activity Readiness Questionnaire for Everyone

Regular physical activity is fun and healthy, and more people should become more physically active every day of the week. Being more physically active is very safe for MOST people. This questionnaire will tell you whether it is necessary for you to seek further advice from your doctor OR a qualified exercise professional before becoming more physically active.

SECTION 1 - GENERAL HEALTH

	Please read the 7 questions below carefully and answer each one honestly: check YES or NO.	YES	NO
1.	Has your doctor ever said that you have a heart condition OR high blood pressure?	○	○
2.	Do you feel pain in your chest at rest, during your daily activities of living, OR when you do physical activity?	○	○
3.	Do you lose balance because of dizziness OR have you lost consciousness in the last 12 months? Please answer NO if your dizziness was associated with over-breathing (including during vigorous exercise).	○	○
4.	Have you ever been diagnosed with another chronic medical condition (other than heart disease or high blood pressure)?	○	○
5.	Are you currently taking prescribed medications for a chronic medical condition?	○	○
6.	Do you have a bone or joint problem that could be made worse by becoming more physically active? Please answer NO if you had a joint problem in the past, but it does not limit your current ability to be physically active. For example, knee, ankle, shoulder or other.	○	○
7.	Has your doctor ever said that you should only do medically supervised physical activity?	○	○

If you answered NO to all of the questions above, you are cleared for physical activity.

Go to Section 3 to sign the form. You do not need to complete Section 2.

- Start becoming much more physically active – start slowly and build up gradually.
- Follow the Canadian Physical Activity Guidelines for your age (www.csep.ca/guidelines).
- You may take part in a health and fitness appraisal.
- If you have any further questions, contact a qualified exercise professional such as a CSEP Certified Exercise Physiologist® (CSEP-CEP) or CSEP Certified Personal Trainer® (CSEP-CPT).
- If you are over the age of 45 yrs. and NOT accustomed to regular vigorous physical activity, please consult a qualified exercise professional (CSEP-CEP) before engaging in maximal effort exercise.

If you answered YES to one or more of the questions above, please GO TO SECTION 2.

Delay becoming more active if:

- You are not feeling well because of a temporary illness such as a cold or fever – wait until you feel better
- You are pregnant – talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the PARmed-X for Pregnancy before becoming more physically active OR
- Your health changes – please answer the questions on Section 2 of this document and/or talk to your doctor or qualified exercise professional (CSEP-CEP or CSEP-CPT) before continuing with any physical activity programme.

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SECTION 2 - CHRONIC MEDICAL CONDITIONS

	Please read the questions below carefully and answer each one honestly: check YES or NO.		YES	NO
1.	Do you have Arthritis, Osteoporosis, or Back Problems?		○ If yes, answer questions 1a-1c	○ If no, go to question 2
	1a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	○	○
	1b.	Do you have joint problems causing pain, a recent fracture or fracture caused by osteoporosis or cancer, displaced vertebra (e.g., spondylolisthesis), and/ or spondylolysis/pars defect (a crack in the bony ring on the back of the spinal column)?	○	○
	1c.	Have you had steroid injections or taken steroid tablets regularly for more than 3 months?	○	○
2.	Do you have Cancer of any kind?		○ If yes, answer questions 2a-2b	○ If no, go to question 3
	2a.	Does your cancer diagnosis include any of the following types: lung/bronchogenic, multiple myeloma (cancer of plasma cells), head, and neck?	○	○
	2b.	Are you currently receiving cancer therapy (such as chemotherapy or radiotherapy)?	○	○
3.	Do you have Heart Disease or Cardiovascular Disease? This includes Coronary Artery Disease, High Blood Pressure, Heart Failure, Diagnosed Abnormality of Heart Rhythm		○ If yes, answer questions 3a-3e	○ If no, go to question 4
	3a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	○	○
	3b.	Do you have an irregular heart beat that requires medical management? (e.g. atrial fibrillation, premature ventricular contraction)	○	○
	3c.	Do you have chronic heart failure?	○	○
	3d.	Do you have a resting blood pressure equal to or greater than 160/90 mmHg with or without medication? (Answer YES if you do not know your resting blood pressure)	○	○
	3e.	Do you have diagnosed coronary artery (cardiovascular) disease and have not participated in regular physical activity in the last 2 months?	○	○
4.	Do you have any Metabolic Conditions? This includes Type 1 Diabetes, Type 2 Diabetes, Pre-Diabetes		○ If yes, answer questions 4a-4c	○ If no, go to question 5
	4a.	Is your blood sugar often above 13.0 mmol/L? (Answer YES if you are not sure)	○	○
	4b.	Do you have any signs or symptoms of diabetes complications such as heart or vascular disease and/or complications affecting your eyes, kidneys, and the sensation in your toes and feet?	○	○
	4c.	Do you have other metabolic conditions (such as thyroid disorders, pregnancy-related diabetes, chronic kidney disease, liver problems)?	○	○
5.	Do you have any Mental Health Problems or Learning Difficulties? This includes Alzheimer's, Dementia, Depression, Anxiety Disorder, Eating Disorder, Psychotic Disorder, Intellectual Disability, Down Syndrome)		○ If yes, answer questions 5a-5b	○ If no, go to question 6
	5a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	○	○
	5b.	Do you also have back problems affecting nerves or muscles?	○	○

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		Please read the questions below carefully and answer each one honestly: check YES or NO.	YES	NO
6.		Do you have a Respiratory Disease? This includes Chronic Obstructive Pulmonary Disease, Asthma, Pulmonary High Blood Pressure	◯ If yes, answer questions 6a-6d	◯ If no, go to question 7
	6a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	◯	◯
	6b.	Has your doctor ever said your blood oxygen level is low at rest or during exercise and/or that you require supplemental oxygen therapy?	◯	◯
	6c.	If asthmatic, do you currently have symptoms of chest tightness, wheezing, laboured breathing, consistent cough (more than 2 days/week), or have you used your rescue medication more than twice in the last week?	◯	◯
	6d.	Has your doctor ever said you have high blood pressure in the blood vessels of your lungs?	◯	◯
7.		Do you have a Spinal Cord Injury? This includes Tetraplegia and Paraplegia	◯ If yes, answer questions 7a-7c	◯ If no, go to question 8
	7a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	◯	◯
	7b.	Do you commonly exhibit low resting blood pressure significant enough to cause dizziness, light-headedness, and/or fainting?	◯	◯
	7c.	Has your physician indicated that you exhibit sudden bouts of high blood pressure (known as Autonomic Dysreflexia)?	◯	◯
8.		Have you had a Stroke? This includes Transient Ischemic Attack (TIA) or Cerebrovascular Event	◯ If yes, answer questions 8a-c	◯ If no, go to question 9
	8a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	◯	◯
	8b.	Do you have any impairment in walking or mobility?	◯	◯
	8c.	Have you experienced a stroke or impairment in nerves or muscles in the past 6 months?	◯	◯
9.		Do you have any other medical condition not listed above or do you live with two chronic conditions?	◯ If yes, answer questions 9a-c	◯ If no, read the advice on page 4
	9a.	Have you experienced a blackout, fainted, or lost consciousness as a result of a head injury within the last 12 months OR have you had a diagnosed concussion within the last 12 months?	◯	◯
	9b.	Do you have a medical condition that is not listed (such as epilepsy, neurological conditions, kidney problems)?	◯	◯
	9c.	Do you currently live with two chronic conditions?	◯	◯

Please proceed to Page 4 for recommendations for your current medical condition and sign this document.

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PAR-Q+

If you answered NO to all of the follow-up questions about your medical condition, you are ready to become more physically active:

- It is advised that you consult a qualified exercise professional (e.g., a CSEP-CEP or CSEP-CPT) to help you develop a safe and effective physical activity plan to meet your health needs.
- You are encouraged to start slowly and build up gradually – 20-60 min. of low- to moderate-intensity exercise, 3-5 days per week including aerobic and muscle strengthening exercises.
- As you progress, you should aim to accumulate 150 minutes or more of moderate-intensity physical activity per week.
- If you are over the age of 45 yrs. and NOT accustomed to regular vigorous physical activity, please consult a qualified exercise professional (CSEP-CEP) before engaging in maximal effort exercise.

If you answered YES to one or more of the follow-up questions about your medical condition:

- You should seek further information from a licensed health care professional before becoming more physically active or engaging in a fitness appraisal and/or visit a or qualified exercise professional (CSEP-CEP) for further information.

Delay becoming more active if:

- You are not feeling well because of a temporary illness such as a cold or fever – wait until you feel better
- You are pregnant - talk to your health care practitioner, your physician, a qualified exercise profesional, and/or complete the PARmed-X for Pregnancy before becoming more physically active OR
- Your health changes - please talk to your doctor or qualified exercise professional (CSEP-CEP) before continuing with any physical activity programme.

SECTION 3 - DECLARATION

- You are encouraged to photocopy the PAR-Q+. You must use the entire questionnaire and NO changes are permitted.
- The Canadian Society for Exercise Physiology, the PAR-Q+ Collaboration, and their agents assume no liability for persons who undertake physical activity. If in doubt after completing the questionnaire, consult your doctor prior to physical activity.
- If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.
- Please read and sign the declaration below:

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that a Trustee (such as my employer, community/fitness centre, health care provider, or other designate) may retain a copy of this form for their records. In these instances, the Trustee will be required to adhere to local, national, and international guidelines regarding the storage of personal health information ensuring that they maintain the privacy of the information and do not misuse or wrongfully disclose such information.

NAME ______________________________ DATE ______________________________

SIGNATURE ______________________________ WITNESS ______________________________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ______________________________

For more information, please contact:
Canadian Society for Exercise Physiology
www.csep.ca

KEY REFERENCES

1. Jamnik VJ, Warburton DER, Makarski J, McKenzie DC, Shephard RJ, Stone J, and Gledhill N. Enhancing the eectiveness of clearance for physical activity participation; background and overall process. APNM 36(S1):S3-S13, 2011.
2. Warburton DER, Gledhill N, Jamnik VK, Bredin SSD, McKenzie DC, Stone J, Charlesworth S, and Shephard RJ. Evidence-based risk assessment and recommendations for physical activity clearance; Consensus Document. APNM 36(S1):S266-s298, 2011.

The PAR-Q+ was created using the evidence-based AGREE process (1) by the PAR-Q+Collaboration chaired by Dr. Darren E. R. Warburton with Dr. Norman Gledhill, Dr. Veronica Jamnik, and Dr. Donald C. McKenzie (2). Production of this document has been made possible through financial contributions from the Public Health Agency of Canada and the BC Ministry of Health Services. The views expressed herein do not necessarily represent the views of the Public Health Agency of Canada or BC Ministry of Health Services.

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MRI Screen

York MRI Facility

Magnetic Resonance (MR) Safety Screening Form

Name ______________________ **Height** ________
Last *First* *Middle* **Weight** ________

Date of Birth ____ / ____ / ____
Month *Day* *Year*

☐ Male ☐ Female ☐ Other

Do you have a **Cardiac Pacemaker** or **Implantable Cardioverter Defibrillator** (ICD)?	☐ Yes	☐ No
Do you have an **Aneurysm Clip**?	☐ Yes	☐ No
Are you Claustrophobic?	☐ Yes	☐ No
Are you currently taking any medications? List: ____________		
Have you ever had an injury involving a metallic object or fragment?	☐ Yes	☐ No
Have you ever worked in a metal shop?	☐ Yes	☐ No
Possibility of pregnancy? ☐ Not Applicable	☐ Yes	☐ No

Surgery

Brain/Head Surgery? ☐ Yes ☐ No
Type/Date: ____________

Heart/Chest Surgery? ☐ Yes ☐ No
Type/Date: ____________

Eye/Ear Surgery? ☐ Yes ☐ No
Type/Date: ____________

Other Surgery? ☐ Yes ☐ No
Type/Date: ____________

Artificial Implants/Mechanical Devices? ☐ Yes ☐ No
Type/Date:

Please Indicate if you have any of the following

Piercings (ear or body)	☐ Yes	☐ No	Medication Patch	☐ Yes	☐ No
Hearing Aid or Cochlear Implant	☐ Yes	☐ No	Tattoo or Permanent Makeup	☐ Yes	☐ No
Permanent Retainer/Braces	☐ Yes	☐ No	Stent or Filter	☐ Yes	☐ No
Dentures or Partials	☐ Yes	☐ No	Antimicrobial Clothing	☐ Yes	☐ No
History of Bullets/Shrapnel/BBs	☐ Yes	☐ No	Intrauterine device (IUD)	☐ Yes	☐ No
History of Seizures	☐ Yes	☐ No	Underwire bra	☐ Yes	☐ No
Hair piece/wig/hair extensions	☐ Yes	☐ No	Magnetic Eyelash Extensions	☐ Yes	☐ No

WARNING: Certain implants, devices, or objects may be hazardous to you and/or may interfere with the MR procedure (i.e., MRI, MR angiography, functional MRI, MR spectroscopy). **Do not enter** the MR magnet room or MR environment if you have any questions or concerns regarding an implant, device, or object. Consult the MRI Technologist or Researcher BEFORE entering the MR system room. **The MR system magnet is ALWAYS on.**

I attest that the above information is correct to the best of my knowledge. I have read and understood the contents of this form and had the opportunity to ask questions regarding the information on this form and regarding the MRI procedure that I am about to undergo.

Signature of person completing form: ____________ **Date** ____ / ____ / ____
M *D* *Y*

Form completed by: ☐ MRI Participant ☐ Other (specify) ____________

Reviewed By: ____________ **PI of study** ____________

Appendix C: PA Questionnaires

International Physical Activity Questionnaire – Short (IPAQ-SF)

INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE

We are interested in finding out about the kinds of physical activities that people do as part of their everyday lives. The questions will ask you about the time you spent being physically active in the **last 7 days**. Please answer each question even if you do not consider yourself to be an active person. Please think about the activities you do at work, as part of your house and yard work, to get from place to place, and in your spare time for recreation, exercise or sport.

Think about all the **vigorous** activities that you did in the **last 7 days**. **Vigorous** physical activities refer to activities that take hard physical effort and make you breathe much harder than normal. Think *only* about those physical activities that you did for at least 10 minutes at a time.

1. During the **last 7 days**, on how many days did you do **vigorous** physical activities like heavy lifting, digging, aerobics, or fast bicycling?

    _____ **days per week**

    ☐ No vigorous physical activities → ***Skip to question 3***

2. How much time did you usually spend doing **vigorous** physical activities on one of those days?

    _____ **hours per day**

    _____ **minutes per day**

    ☐ Don’t know/Not sure

Think about all the **moderate** activities that you did in the **last 7 days**. **Moderate** activities refer to activities that take moderate physical effort and make you breathe somewhat harder than normal. Think only about those physical activities that you did for at least 10 minutes at a time.

3. During the **last 7 days**, on how many days did you do **moderate** physical activities like carrying light loads, bicycling at a regular pace, or doubles tennis? Do not include walking.

    _____ **days per week**

    ☐ No moderate physical activities → ***Skip to question 5***

4. How much time did you usually spend doing **moderate** physical activities on one of those days?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

Think about the time you spent **walking** in the **last 7 days**. This includes at work and at home, walking to travel from place to place, and any other walking that you have done solely for recreation, sport, exercise, or leisure.

5. During the **last 7 days**, on how many days did you **walk** for at least 10 minutes at a time?

_____ **days per week**

☐ No walking → ***Skip to question 7***

6. How much time did you usually spend **walking** on one of those days?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

The last question is about the time you spent **sitting** on weekdays during the **last 7 days**. Include time spent at work, at home, while doing course work and during leisure time. This may include time spent sitting at a desk, visiting friends, reading, or sitting or lying down to watch television.

7. During the **last 7 days**, how much time did you spend **sitting** on a **week day**?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

This is the end of the questionnaire, thank you for participating.

HUNT PA Questionnaire

HUNT 1 https://www.ntnu.edu/c/document_library/get_file?uuid=a173dabd-d59e-4be1-ad40-fcd1b9
0

Exercise: going for walks, skiing, swimming or training / sport

1 How frequently do you exercise? (on average)

- ☐ Never
- ☐ Less than once a week
- ☐ Once a week
- ☐ 2-3 times per week
- ☐ Almost every day

2 If you do exercise as frequently as once or more per week, how hard do you push yourself?

- ☐ I take it easy without breaking into a sweat or losing my breath
- ☐ I push myself so hard that I lose my breath and break into a sweat
- ☐ I push myself to near-exhaustion

3 How long does each session last? (on average)

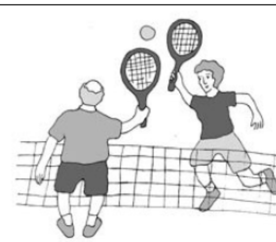
- ☐ Less than 15 minutes
- ☐ 16-30 minutes
- ☐ 31 minutes to 1 hour
- ☐ More than 1 hour

Rapid Assessment of Physical Activity

Physical Activities are activities where you move and increase your heart rate above its resting rate, whether you do them for pleasure, work, or transportation.

The following questions ask about the amount and intensity of physical activity you usually do. The intensity of the activity is related to the amount of energy you use to do these activities.

Examples of physical activity intensity levels:

Light activities • your heart beats slightly faster than normal • you can talk and sing	 Walking Leisurely  Stretching  Vacuuming or Light Yard Work
Moderate activities • your heart beats faster than normal • you can talk but not sing	 Fast Walking  Aerobics Class  Strength Training  Swimming Gently
Vigorous activities • your heart rate increases a lot • you can't talk or your talking is broken up by large breaths	 Stair Machine  Jogging or Running  Tennis, Racquetball, Pickleball or Badminton

How physically active are you? *(Check one answer on each line)*

			Does this accurately describe you?	
RAPA 1	1	I rarely or never do any physical activities.	Yes ☐	No ☐
	2	I do some **light** or **moderate** physical activities, but not every week.	Yes ☐	No ☐
	3	I do some **light** physical activity every week.	Yes ☐	No ☐
	4	I do **moderate** physical activities every week, but less than 30 minutes a day or 5 days a week.	Yes ☐	No ☐
	5	I do **vigorous** physical activities every week, but less than 20 minutes a day or 3 days a week.	Yes ☐	No ☐
	6	I do 30 minutes or more a day of **moderate** physical activities, 5 or more days a week.	Yes ☐	No ☐
	7	I do 20 minutes or more a day of **vigorous** physical activities, 3 or more days a week.	Yes ☐	No ☐
RAPA 2 (3 = Both 1 & 2)	1	I do activities to increase muscle **strength**, such as lifting weights or calisthenics, once a week or more.	Yes ☐	No ☐
	2	I do activities to improve **flexibility**, such as stretching or yoga, once a week or more.	Yes ☐	No ☐

ID # ______________________________

Today's Date ______________________

RAND 36-Item Health Survey 1.0 Questionnaire Items

1. In general, would you say your health is:	
Excellent	1
Very good	2
Good	3
Fair	4
Poor	5

2**. Compared to one year ago,** how would your rate your health in general **now**?	
Much better now than one year ago	1
Somewhat better now than one year ago	2
About the same	3
Somewhat worse now than one year ago	4
Much worse now than one year ago	5

Note: The WSIB acknowledges that the RAND 36-Item Short Form Health Survey was developed at RAND as part of the Medical Outcomes Study.

The following items are about activities you might do during a typical day. Does **your health now limit you** in these activities? If so, how much?

(Circle One Number on Each Line)

	Yes, Limited a Lot	Yes, Limited a Little	No, Not limited at All
3. **Vigorous activities,** such as running, lifting heavy objects, participating in strenuous sports	[1]	[2]	[3]
4. **Moderate activities,** such as moving a table, pushing a vacuum cleaner, bowling, or playing golf	[1]	[2]	[3]
5. Lifting or carrying groceries	[1]	[2]	[3]
6. Climbing **several** flights of stairs	[1]	[2]	[3]
7. Climbing **one** flight of stairs	[1]	[2]	[3]
8. Bending, kneeling, or stooping	[1]	[2]	[3]
9. Walking **more than a mile**	[1]	[2]	[3]
10. Walking **several blocks**	[1]	[2]	[3]
11. Walking **one block**	[1]	[2]	[3]
12. Bathing or dressing myself	[1]	[2]	[3]

Note: The WSIB acknowledges that the RAND 36-Item Short Form Health Survey was developed at RAND as part of the Medical Outcomes Study.

During the **past 4 weeks**, have you had any of the following problems with your work or other regular daily activities **as a result of your physical health**?

(Circle One Number on Each Line)

	Yes	No
13. Cut down the amount of time you spent on work or other activities	1	2
14. **Accomplished less** than you would like	1	2
15. Were limited in the **kind** of work or other activities	1	2
16. Had **difficulty** performing the work or other activities (for example, it took extra effort)	1	2

During the **past 4 weeks**, have you had any of the following problems with your work or other regular daily activities **as a result of any emotional problems** (such as feeling depressed or anxious)?

(Circle One Number on Each Line)

	Yes	No
17. Cut down the **amount of time** you spent on work or other activities	1	2
18. **Accomplished less** than you would like	1	2
19. Didn't do work or other activities as **carefully** as usual	1	2

Note: The WSIB acknowledges that the RAND 36-Item Short Form Health Survey was developed at RAND as part of the Medical Outcomes Study.

20. During the **past 4 weeks**, to what extent has your physical health or emotional problems interfered with your normal social activities with family, friends, neighbours, or groups?

(Circle One Number)

Not at all 1

Slightly 2

Moderately 3

Quite a bit 4

Extremely 5

21. How much **bodily** pain have you had during the **past 4 weeks**?

(Circle One Number)

None 1

Very mild 2

Mild 3

Moderate 4

Severe 5

Very severe 6

22. During the **past 4 weeks**, how much did **pain** interfere with your normal work (including both work outside the home and housework)?

(Circle One Number)

Not at all 1

A little bit 2

Moderately 3

Quite a bit 4

Extremely 5

Note: The WSIB acknowledges that the RAND 36-Item Short Form Health Survey was developed at RAND as part of the Medical Outcomes Study.

These questions are about how you feel and how things have been with you **during the past 4 weeks.** For each question, please give the one answer that comes closest to the way you have been feeling.
How much of the time during the past 4 weeks . . .

(Circle One Number on Each Line)

	All of the Time	Most of the Time	A Good Bit of the Time	Some of the Time	A Little of the Time	None of the Time
23. Did you feel full of pep?	1	2	3	4	5	6
24. Have you been a very nervous person?	1	2	3	4	5	6
25. Have you felt so down in the dumps that nothing could cheer you up?	1	2	3	4	5	6
26. Have you felt calm and peaceful?	1	2	3	4	5	6
27. Did you have a lot of energy?	1	2	3	4	5	6
28. Have you felt downhearted and blue?	1	2	3	4	5	6
29. Did you feel worn out?	1	2	3	4	5	6
30. Have you been a happy person?	1	2	3	4	5	6
31. Did you feel tired?	1	2	3	4	5	6

Note: The WSIB acknowledges that the RAND 36-Item Short Form Health Survey was developed at RAND as part of the Medical Outcomes Study.

32. During the **past 4 weeks**, how much of the time has your **physical health or emotional problems** interfered with your social activities (like visiting with friends, relatives, etc.)?

(Circle One Number)

All of the time 1

Most of the time 2

Some of the time 3

A little of the time 4

None of the time 5

How **TRUE** or **FALSE** is each of the following statements for you.

(Circle One Number on Each Line)

	Definitely True	Mostly True	Don't Know	Mostly False	Definitely False
33. I seem to get sick a little easier than other people	1	2	3	4	5
34. I am as healthy as anybody I know	1	2	3	4	5
35. I expect my health to get worse	1	2	3	4	5
36. My health is excellent	1	2	3	4	5

Note: The WSIB acknowledges that the RAND 36-Item Short Form Health Survey was developed at RAND as part of the Medical Outcomes Study.

Appendix D: Custom Questionnaires

Intake Form

Power of Exercise - Intake Form

Today's Date:	mmm/dd/yyyy	Time of Collection:	hh:mm am/pm
Participant Code:			

Participant Information

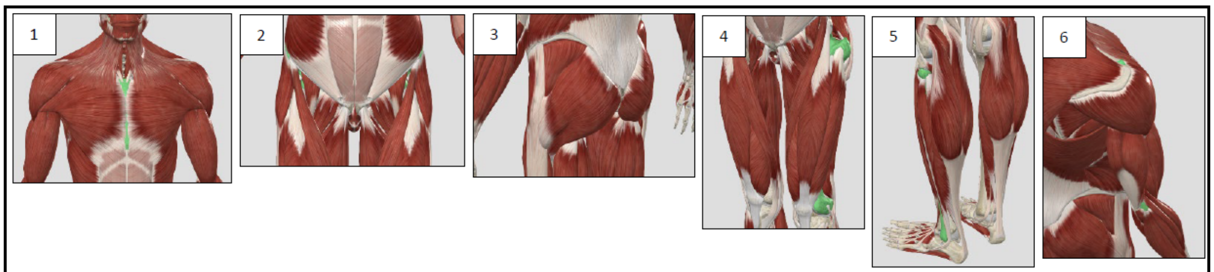
Name:			
DOB:	mmm/dd/yyyy		
Sex (at birth):		Gender:	
Masters Athlete*:	☐ Yes ☐ No	Athlete**:	☐ Yes ☐ No
Primary Sport:			

* Masters athletes are defined as "individuals who systematically train for, and compete in, organized forms of competitive sport specifically designed for older adults" (Geard et al., 2017).

** Athletes are defined as individuals "who participate in an organized team or individual sport that requires regular competition against others as a central component, places a high premium on excellence and achievement, and requires some form of systematic (and usually intense) training" (Maron et al., 2015).

Anthropometrics

Height (m):			Weight (kg):		
Body Fat (%):			Handgrip Strength	Left:	Right:
Chest @ nipple (cm):			Waist @ naval (cm):		
Hips @ widest (cm):			Thigh @ 50% from GT to lat condyle (cm):	Left:	Right:
Shank @ 50% from fib head to lat mall (cm):	Left:	Right:	Arm (cm):	Left:	Right:



1) Chest circumference @ nipple line
2) Waist circumference @ naval
3) Hip circumference @ largest girth
4) Thigh circumference @ 70% of distance moving proximally between greater trochanter and lateral epicondyle of femur
5) Shank circumference @ 66% of distance moving proximally between lateral malleolus and fibular head
6) Arm circumference @ 50% of distance between a/c joint and lateral epicondyle of humerus

Custom Questionnaire (Injury and PA History)

Power of Exercise - Custom Survey

Section 1. This section deals with **repetitive strain injuries**. By this we mean injuries caused by overuse or by repeating the same movement frequently. For example, carpal tunnel syndrome, tennis elbow or tendonitis.

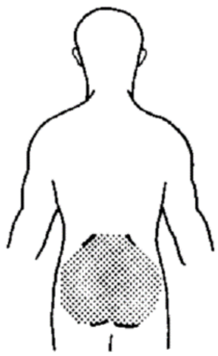
1. In the past 12 months, did you experience any injuries due to repetitive strain which were serious enough to limit your normal activities?
   a) Yes
   b) No (skip to Section 2)
2. Thinking about the most serious repetitive strain, what part of the body was affected?
   a) Head
   b) Neck
   c) Shoulder, upper arm
   d) Elbow, lower arm
   e) Wrist
   f) Hand
   g) Hip
   h) Thigh
   i) Knee, lower leg
   j) Ankle, foot
   k) Upper back or upper spine (excluding neck)
   l) Lower back or lower spine
   m) Chest (excluding back and spine)
   n) Abdomen or pelvis (excluding back and spine)

Power of Exercise - Custom Survey

Section 2. Now some questions about **acute injuries** which occurred in the past 12 months, and were serious enough to limit your normal activities. For example, a broken bone, a bad cut or burn, a sprain, or a poisoning.

3. Not counting repetitive strain injuries, were you injured in the past 12 months?
   a) Yes
   b) No (skip to Section 3)
4. How many times were you injured?
   ____
5. Thinking about the most serious injury, what type of injury did you have?
   a) Multiple injuries
   b) Broken or fractured bones
   c) Burn, scald, chemical burn
   d) Dislocation
   e) Sprain or strain
   f) Cut, puncture, animal or human bite (open wound)
   g) Scrape, bruise, blister
   h) Concussion or other brain injury
   i) Poisoning
   j) Injury to internal organs
6. Thinking of this same injury, what part of the body was injured?
   a) Head
   b) Neck
   c) Shoulder, upper arm
   d) Elbow, lower arm
   e) Wrist
   f) Hand
   g) Hip
   h) Thigh
   i) Knee, lower leg
   j) Ankle, foot
   k) Upper back or upper spine (excluding neck)
   l) Lower back or lower spine
   m) Chest (excluding back and spine)
   n) Abdomen or pelvis (excluding back and spine)

Section 3. Complete the following *only if* you indicated overuse or acute **injury of the low back** (question 2 L or 6 L). Adapted from Bahr et al., 2004, Fett et al., 2017, and Kuorinka & Andersson 1987.



LOW BACK

How to answer the questionnaire: In this picture you can see the approximate position of the part of the body referred to in the questionnaire. By low back trouble is meant ache, pain or discomfort in the shaded area whether or not it extends from there to one or both legs (sciatica).

Please answer by putting a cross in the appropriate box — one cross for each question. You may be in doubt as to how to answer, but please do your best anyway.

7. Have you been examined or treated for LBP by a physician, physical therapist, chiropractor, or other health personnel as an outpatient during the previous 12 months?
   a) Yes
   b) No
8. Have you ever been admitted to hospital because of LBP?
   a) Yes
   b) No
9. Have you ever had surgery because of LBP?
   a) Yes
   b) No
10. Have you ever had to change your occupation or working assignments because of LBP?
    a) Yes
    b) No
11. Have you experienced LBP during the previous 7 days?
    a) Yes
    b) No
12. Have you ever had radiating LBP?
    a) Yes
    b) No
13. How many days during the past 12 months have you had LBP?

    _______ days

Power of Exercise - Custom Survey

Section 4. Please describe your **physical activity history**. Some examples include casual weight-training, recreational soccer, or varsity basketball. An example is provided in the first row. You may add additional details as you see fit.

Activity	Competitive Level*	Duration (h) per session	Sessions per week	Weeks per year	Started (Year)	Ended (Year)
Volleyball	*High school*	*2*	*3*	*16*	*1982*	*1984*

* List the competitive level if applicable – recreational / non-competitive, varsity, national, international

Power of Exercise - Custom Survey

Section 5. This section collects additional information regarding your occupation (or past occupation) and any related physical activity.

14. What is/was your occupation? ____
15. Does your work involve **vigorous-intensity activity** that causes **large increases in breathing or heart rate** like carrying/lifting heavy loads, digging, or construction work for at least 10 minutes continuously?
    a) Yes
    b) No (skip to Question 18)
16. In a typical week, how many days do you do vigorous-intensity activities as part of your work?
    ______ days
17. How much time do you spend doing vigorous-intensity activities at work on a typical day?
    ______ hours ______ minutes
18. Does your work involve **moderate intensity activity** that causes **small increases in breathing or heart rate** such as brisk walking or carrying light loads for at least 10 minutes continuously?
    a) Yes
    b) No (skip to Question 21)
19. In a typical week, how many days do you do moderate-intensity activities as part of your work?
    ______ days
20. How much time do you spend doing moderate-intensity activities at work on a typical day?
    ______ hours ______ minutes
21. What is your income bracket?
    a) 0-50k
    b) 51-100k
    c) 101-155k
    d) 156-221k
    e) 222k+

Power of Exercise - Custom Survey

Section 6. This section asks about miscellaneous factors that may affect your physical performance.

22. Are you taking any medications that may affect your physical performance? Examples would include lipid lowering drugs (i.e., Lipitor), androgens (i.e., testosterone), ADHD medication (i.e., methylphenidate).
23. What is the date of your last menstrual cycle (if applicable)? ______________________________
24. Have you had any surgeries? If so, please list them below with the approximate date.
    1. ____ Date: mmm/dd/yyyy
    2. ____ Date: mmm/dd/yyyy
    3. ____ Date: mmm/dd/yyyy
    4. ____ Date: mmm/dd/yyyy
    5. ____ Date: mmm/dd/yyyy
25. Have you ever been diagnosed with osteoarthritis? If yes, please list the joint(s).
    ____

Activity Tracking Log

Power of Exercise - Activity Tracking Log

- Please wear your activity monitor on your non-dominant wrist for one full week, 24h per day.
- If you remove it and forget to wear it, please write down any activities including date, time and duration in the log provided below.
- You do not need to remove it when showering or swimming (except in salt water)

Activity	**Date**	**Time**	**Duration (min)**
	mmm/dd/yyyy	hh:mm am/pm	
	mmm/dd/yyyy	hh:mm am/pm	
	mmm/dd/yyyy	hh:mm am/pm	
	mmm/dd/yyyy	hh:mm am/pm	
	mmm/dd/yyyy	hh:mm am/pm	
	mmm/dd/yyyy	hh:mm am/pm	
	mmm/dd/yyyy	hh:mm am/pm	

Exit Questionnaire

Power of Exercise - Initial Exit Survey

1. How did you find the tasks in today's session?

   ____
   ____
   ____

2. Would you recommend participating in this study to your friends?

   ____
   ____
   ____

3. Do you have any questions for us?

   ____
   ____
   ____

4. Is there any information, anecdotes, and/or concerns you would like to share?

   ____
   ____
   ____

Appendix E: MRI DTI Wolfram Mathematica Pipeline

Load QMRITools package and define directory paths

```
In[ ]:= << QMRITools`
        Dir = SetDirectory[NotebookDirectory[]]
```

Convert dicom files to nifti (.nii) format

First, select the folder containing the raw dicom files for the main diffusion sequence. Next, select the folder where you would like to save the nifti format diffusion files.

```
In[ ]:= DcmToNii[];
        Using Chris Rorden's dcm2niix.exe (https://github.com/rordenlab/dcm2niix)
```

Select the folder containing the raw dicom files for the Dixon Water image. Then select the folder where you would like to save the nifti format water file.

```
In[ ]:= DcmToNii[];
        Using Chris Rorden's dcm2niix.exe (https://github.com/rordenlab/dcm2niix)
```

Import the .nii format diffusion, water, and segmentation images into QMRITools

Select the dwi .nii format file and the associated .bval and .bvec files created in the previous step for import.

```
In[ ]:= {dwi, grad, val, voxD} = ImportNiiDiff[];
        ExportNii[dwi, voxD, "dwi.nii"];
```

Select the water nifti format file for import.

```
In[ ]:= {water, voxW} = ImportNii[];
        ExportNii[water, voxW, "water.nii"];
```

Select the segmentation mask nifti file for import.

```
In[ ]:= {labels, voxL} = ImportNii[];
        ExportNii[labels, voxL, "labels.nii"];
```

Check the imported nifti images carefully

Plot the diffusion, water, and segmentation images for visual assessment. This step is an important verification of data quality and content.

```
In[ ]:= PlotData[dwi, voxD];
In[ ]:= PlotData[water, voxW];
In[ ]:= PlotData[labels, water];
```

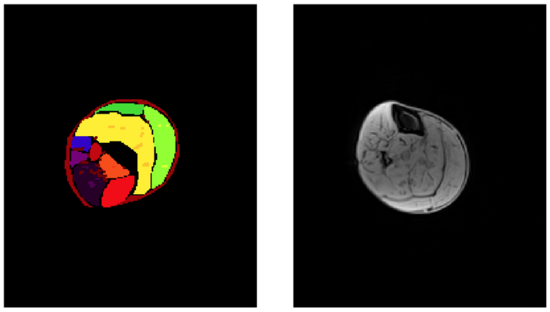
Flip labels image orientation (if necessary)

IMPORTANT If you observe that the segmentation mask image has been flipped by ITK-SNAP and is not in the same orientation as the water image as a result, use the "RotateData" command below to transform the segmentation to the correct orientation. Verify the new orientation by plotting the labels relative to the water image.

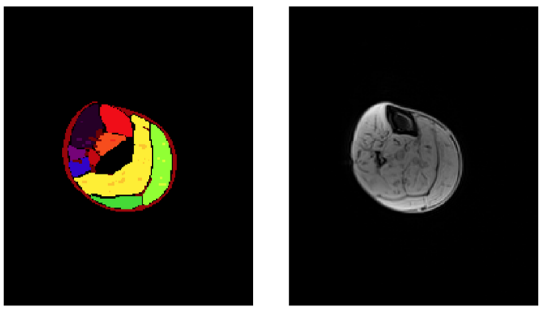
```
In[ ]:= labelsFlip = RotateData[labels[[All]]];
        ExportNii[labelsFlip, voxW, "labelsFlip.nii"];

In[ ]:= PlotData[labelsFlip, water];
```

Before image rotation



After image rotation



Calculate masks for the water and diffusion images

```
In[∘]:= waterMask = Dilation[Mask[water], 5];
PlotData[water, waterMask, voxW];
```

```
In[∘]:= dwiMean = NormalizeMeanData@dwi;
ExportNii[dwiMean, voxD, "dwiMean.nii"]
dwiMask = Mask[dwiMean, 20, MaskSmoothing → True];
ExportNii[dwiMask, voxD, "dwiMask.nii"]
PlotData[dwiMean, dwiMask, voxD]
```

Remove Rician/thermal noise using PCA via PCADeNoise - *3 min*

Output of this step is a denoised diffusion dataset called “dwiN.nii.gz”.

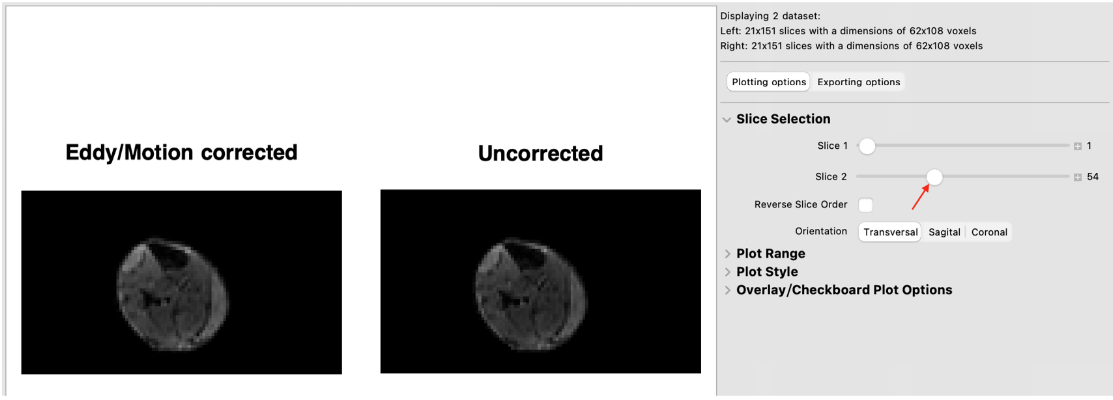
```
In[∘]:= {dwiN, sig} = PCADeNoise[MaskData[dwi, dwiMask]];
ExportNii[dwiN, voxD, "dwiN.nii"];
PlotData[dwi, dwiN, voxD];
```

Perform motion and eddy current correction - *5 min*

This step takes the denoised diffusion data created in the previous step and performs motion and eddy current correction. Output of this step is the file “dwiReg.nii.gz”.

```
In[∘]:= dwiReg = RegisterDiffusionData[{dwiN[[All]], dwiMask, voxD}];
ExportNii[dwiReg, voxD, "dwiReg.nii"];
PlotData[dwiReg, dwiN, voxD];
```

Visually compare the registered diffusion data with non-registered data. Use the “Slice 2” slider to scan through all volumes of the images to verify that motion has been reduced in the corrected image.



Perform IVIM correction (remove perfusion effects)

Calculate the geometric mean of the diffusion data for each unique b-value. Calculate the IVIM fit and perform correction. Output is "dwiRegIVIM.nii.gz".

```
In[ ]:= {mean, valU} = MeanBvalueSignal[dwiReg, val];
fitIVIM = IVIMCalc[MeanSignal[mean], valU, {1, .05, .003, .015}, IVIMFixed → True];
{s0i, fri, adci, pD} =
  Quiet@IVIMCalc[mean, valU, fitIVIM, IVIMConstrained → False, Parallelize → True, MonitorIVIMCalc → False,
    IVIMFixed → True];
dwiRegIVIM = First@IVIMCorrectData[dwiReg, {s0i, Clip[fri, {0, 1}, {0, 1}], pD}, val, FilterMaps → False];
ExportNii[dwiRegIVIM, voxD, "dwiRegIVIM.nii"];
PlotData[dwiRegIVIM, voxD];
```

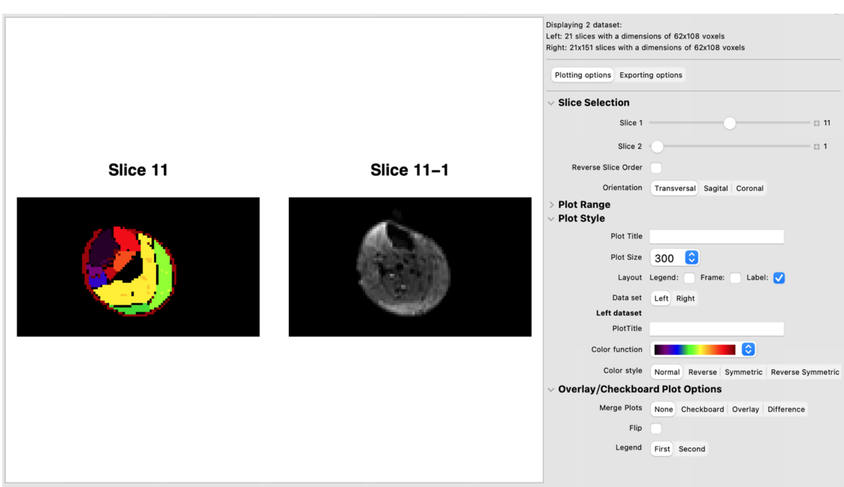
Align the labels image to the diffusion data

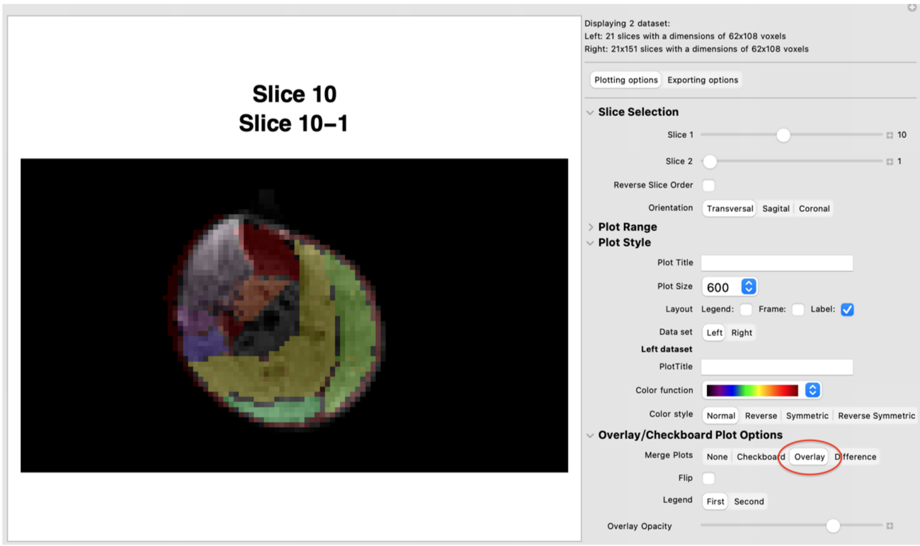
This step is necessary to ensure that the segmentation mask labels (which are in anatomical space) are spatially aligned with the diffusion data.

```
In[ ]:= {waterDWI, labelsFlipDWI} = RegisterDataTransform[{dwiMean, dwiMask, voxD}, {water, waterMask, voxW},
   {labelsFlip, voxL}, Iterations → 250, MethodReg → {"rigid", "bspline"}, TransformMethod → "Segmentation"];
ExportNii[waterDWI, voxD, "waterDWI.nii"]
ExportNii[labelsFlipDWI, voxD, "labelsFlipDWI.nii"]
```

As illustrated below, overlay the segmentation labels and the diffusion images by clicking "Overlay" in the image viewer. Use the "Overlay Opacity" slider to switch between the diffusion and label images to verify that they are properly aligned.

```
In[ ]:= PlotData[labelsFlipDWI, dwiRegIVIM]
```





Calculate DTI metrics - *3 min*

```
In[∘]:= {tens, s0, out, res} =
          Quiet@TensorCalc[MaskData[dwiRegIVIM, dwiMask], grad, val, FullOutput → True, Method → "iWLLS",
            RobustFit → True, Parallelize → True, MonitorCalc → False];
        ExportNii[Transpose@tens, voxD, "tens.nii"];
        PlotData[Transpose@tens, res, voxD];
        {l1, l2, l3, md, fa} = ParameterCalc[tens];
        PlotData[md, fa, voxD, PlotRange → {{0, 3}, {0, 1}}];
        ExportNii[md, voxD, "md.nii"];
        ExportNii[fa, voxD, "fa.nii"];
        ExportNii[l1, voxD, "l1.nii"];
        ExportNii[l2, voxD, "l2.nii"];
        ExportNii[l3, voxD, "l3.nii"];
        pd = l1;
        ExportNii[pd, voxD, "pd.nii"];
        rd = Mean[{l2, l3}];
        ExportNii[rd, voxD, "rd.nii"];
```

Extract mean FA, MD, PD, and RD from segmented regions

Calculates the mean, standard deviation, median, 5% and 95% confidence intervals for each segmentation label. Values are exported in "FAvalues.csv" , "MDvalues.csv", "PDvalues.csv", and "RDvalues.csv" files.

```
In[∘]:= {masks, labs} = SplitSegmentations[labelsFlipDWI];
        ExportNii[masks, voxD, "masks.nii"];
        FAvalues = TableForm@GetMaskMeans[fa, masks, "FA"]
        Export["FAvalues.csv", FAvalues];
        MDvalues = TableForm@GetMaskMeans[md, masks, "MD"]
        Export["MDvalues.csv", MDvalues];
        PDvalues = TableForm@GetMaskMeans[pd, masks, "PD"]
        Export["PDvalues.csv", PDvalues];
        RDvalues = TableForm@GetMaskMeans[rd, masks, "RD"]
        Export["RDvalues.csv", RDvalues];
```

Out[∘]//TableForm=

FA Mean	FA Std	FA Median	FA 5%	FA 95%
0.278514	0.0265984	0.279315	0.233421	0.320853
0.242209	0.0204445	0.242209	0.208581	0.275837
0.251163	0.0358051	0.251163	0.192269	0.310057
0.229944	0.0479949	0.229944	0.150999	0.308888
0.247253	0.0260155	0.247253	0.204461	0.290045
0.217062	0.0332692	0.216241	0.163784	0.273162
0.187508	0.0217845	0.189416	0.148613	0.219928
0.1993	0.0486342	0.198222	0.121199	0.281105
0.195065	0.0364636	0.193763	0.137385	0.257223
0.26992	0.0460477	0.26992	0.194178	0.345661
0.	0.	0.	0.	0.
0.690894	0.395781	0.611018	0.207244	1.45248
0.266748	0.131292	0.266748	0.0507911	0.482704
0.235467	0.0688177	0.235467	0.122272	0.348662
0.263327	0.0589195	0.263327	0.166413	0.360241

Out[∘]//TableForm=

MD Mean	MD Std	MD Median	MD 5%	MD 95%
1.52506	0.0921194	1.52506	1.37353	1.67658
1.51858	0.0761762	1.51858	1.39329	1.64388
1.50713	0.147009	1.5048	1.2694	1.75284
1.52954	0.111019	1.52196	1.36042	1.7246
1.61617	0.104899	1.61617	1.44363	1.78871
1.63493	0.115131	1.63577	1.44414	1.82283
1.68582	0.053747	1.68354	1.60144	1.77804
1.45788	0.135559	1.45454	1.24078	1.68645
1.37451	0.0916032	1.37451	1.22383	1.52518
1.46912	0.150886	1.45579	1.24478	1.73869
0.	0.	0.	0.	0.
0.734323	0.90237	0.564657	−0.410918	2.45587
1.45555	0.616256	1.45555	0.441895	2.4692
1.63711	0.296	1.60173	1.21432	2.17823
1.49244	0.201918	1.49746	1.15188	1.81572

Out[∘]//TableForm=

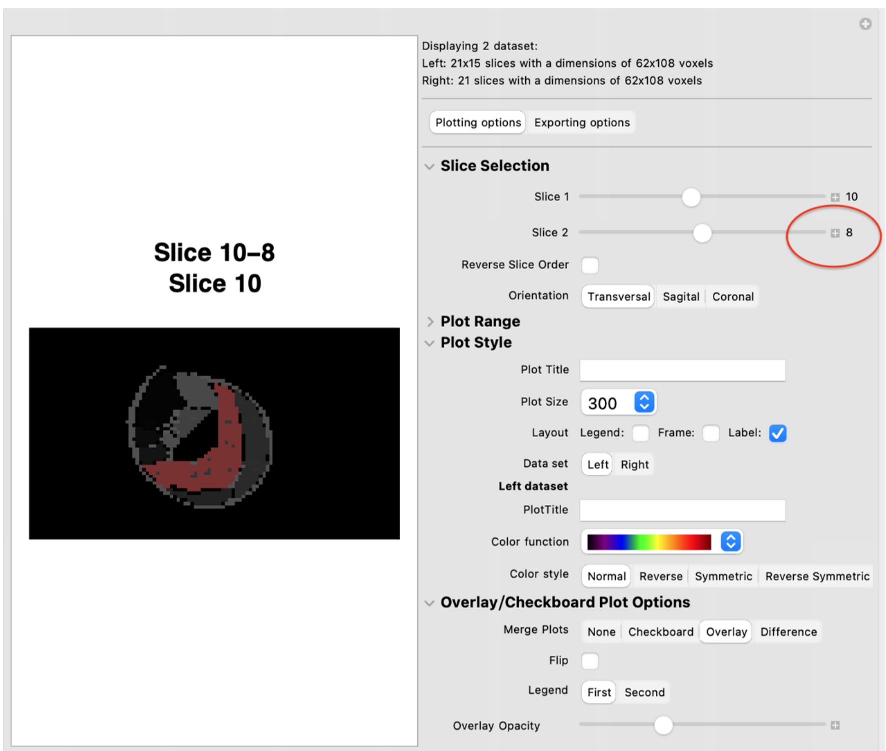
PD Mean	PD Std	PD Median	PD 5%	PD 95%
2.00507	0.103159	2.00507	1.83539	2.17475
1.9727	0.0537782	1.96927	1.89035	2.06682
1.91043	0.166379	1.9371	1.59829	2.134
1.91115	0.0945384	1.91115	1.75565	2.06665
2.06646	0.117728	2.0701	1.86671	2.25368
2.02446	0.150103	2.04041	1.75261	2.24264
2.03553	0.0408896	2.04062	1.96053	2.09354
1.78473	0.1358	1.77646	1.57605	2.02177
1.74741	0.144323	1.71789	1.57145	2.0256
1.86989	0.176343	1.86989	1.57983	2.15995
0.	0.	0.	0.	0.
0.747942	1.00149	0.747942	−0.899369	2.39525
1.40926	0.804519	1.57016	−0.137063	2.39717
1.87103	0.321998	1.89907	1.29634	2.35053
1.8825	0.207336	1.89701	1.51767	2.19769

Out[∘]//TableForm=

RD Mean	RD Std	RD Median	RD 5%	RD 95%
1.27972	0.0940774	1.27972	1.12498	1.43446
1.28903	0.0952086	1.28903	1.13243	1.44564
1.27855	0.165829	1.27855	1.00578	1.55131
1.33447	0.151957	1.31968	1.11105	1.60784
1.38161	0.107322	1.38161	1.20508	1.55814
1.43122	0.11066	1.43122	1.2492	1.61324
1.49591	0.0331985	1.49043	1.45167	1.55837
1.29696	0.149894	1.2923	1.05862	1.55132
1.29646	0.130524	1.27004	1.13716	1.54772
1.25517	0.136942	1.2399	1.05747	1.50412
0.	0.	0.	0.	0.
0.536206	0.856477	0.390993	−0.595504	2.15201
1.2373	0.71597	1.2373	0.0596335	2.41497
1.36272	0.290933	1.36272	0.884174	1.84126
1.32906	0.213711	1.30284	1.0251	1.72058

You can plot the individual masks in the segmentation file to verify which table row belongs with which region. If needed, overlay the individual mask images onto the whole-leg segmentation image and adjust the "Overlay Opacity" slider so that you can see both images. You can also change the "Color Function" to see the overlaid mask better. Scroll through the "Slice 2" values to see which row number corresponds with each region.

In[ ]:= `PlotData[masks, labelsFlipDWI]`



Appendix F: Calculation of Diffusion Parameters

Diffusion tensor computation yields three eigenvalues λ_1, λ_2 & λ_3, where λ_1 is the first and largest eigenvalue. In the cylindrical cell body of muscle fiber, λ_1 corresponds to a measure of diffusion strength in the axial direction and to the parameter PD (parallel diffusion). Eigenvalues λ_2 & λ_3 correspond to diffusion strength in the radial direction of the muscle fiber and the diffusion parameter RD (radial diffusion) is simply an average of these two eigenvalues. Based on these three eigenvalues we can compute the remaining standard DTI parameters MD and FA. MD, or mean diffusion (**Equation E.1**), describes the average diffusion strength in all three directions, and FA, or fractional anisotropy (**Equation E.2**), quantifies the extent to which diffusion is focused in one direction. A value of zero implies fully isotropic diffusion (random in all directions) and a value of one implies maximally anisotropic diffusion (diffusion in one direction only).

Equation E.1 $$MD = \frac{(\lambda_1+\lambda_2+\lambda_3)}{3}$$

Equation E.2 $$FA = \sqrt{\frac{1}{2}} \cdot \frac{\sqrt{(\lambda_1-\lambda_2)^2+(\lambda_2-\lambda_3)^2+(\lambda_3-\lambda_1)^2}}{\sqrt{\lambda_1+\lambda_2+\lambda_3}}$$

In muscle cells, the larger fiber diameter of Type II fibers is theorized to yield higher radial diffusion (RD) (Scheel et al., 2013). Similarly, the microstructure within Type I fibers is theorized to yield lower diffusion in the radial direction (RD) and this higher FA. One possible mechanism for this is the inhibition of diffusion by sarcoplasmic reticulum and mitochondrial microstructures as discussed in (Scheel et al., 2013).

Appendix G: Single Versus Dual Force Plate

A methodological error was discovered at participant 38, affecting all previous participants. The error involved the use of a single force plate on the lower platform which was used to measure joint moments and powers during the CMJ and chair rise tasks. Lower-body power measured as the product of net force and vertical velocity of the pelvis centre of mass, which is the primary independent variable in all research questions, was *not affected* by this error. Joint powers presented as part of Research Question #1 (Table 4.1 and 4.2) were affected by this error and therefore this analysis was conducted prior to presenting those results.

It is not good practice to use one force plate to compute inverse dynamics because a single force plate yields only one centre of pressure (CoP) which is the location where the ground reaction force (GRF) is applied to the model. Having two feet in contact with the same platform is problematic because the second foot acts as a redundant support making the inverse dynamics equations mathematically unsolvable. Visually speaking (**Figure G1**), on a single force plate during quiet standing, the CoP would likely be somewhere in between both feet (viewed in the frontal plane) however there is no direct way to quantify this symmetry from single force plate kinetic data. Kinematic data can be used to assess the CoP symmetry by comparing the distance from the CoP to the left and right heel, respectively, at the instant of peak GRF. Participants 39 onward were evaluated using two force plates, and thus the symmetry of their GRFs can be compared.

Perhaps the simplest and most transparent remedy for conducting inverse dynamics on the first 38 participants is to assume GRF symmetry and then evaluate that assumption in the subsequent 39 participants. This assumption implies that the first 38 participants had 50% of the total GRF applied to the same sagittal plane location of both feet. Only the sagittal plane is of concern in this case since only sagittal joint moments and joint powers are studied in this dissertation. Both analyses are provided below.

CoP Symmetry

Symmetry of the CoP was calculated with the following script in Visual3D:

```
1    Evaluate_Expression
2    /EXPRESSION= (COFP::ORIGINAL::FP1::X - KINETIC_KINEMATIC::LFT::CGPos::X) /
     (KINETIC_KINEMATIC::RFT_2::CGPos::X - KINETIC_KINEMATIC::LFT::CGPos::X)
3    /RESULT_TYPES=DERIVED
4    /RESULT_FOLDERS=PROCESSED
5    /RESULT_NAME=COP_BALANCE
6    ;
```

Line 2 solves for the distance between the CoP (along the x-axis) and the left heel (CoP X minus LFT), then divides that by the distance between both heels (RFT – LFT; **Figure G2** left) yielding a ratio (Left/Total; **Figure G2** right). The resulting ratio describes the location of the CoP relative to the left heel with a value of 0.0 being directly under the left heel and 1.0 being directly under the right heel.

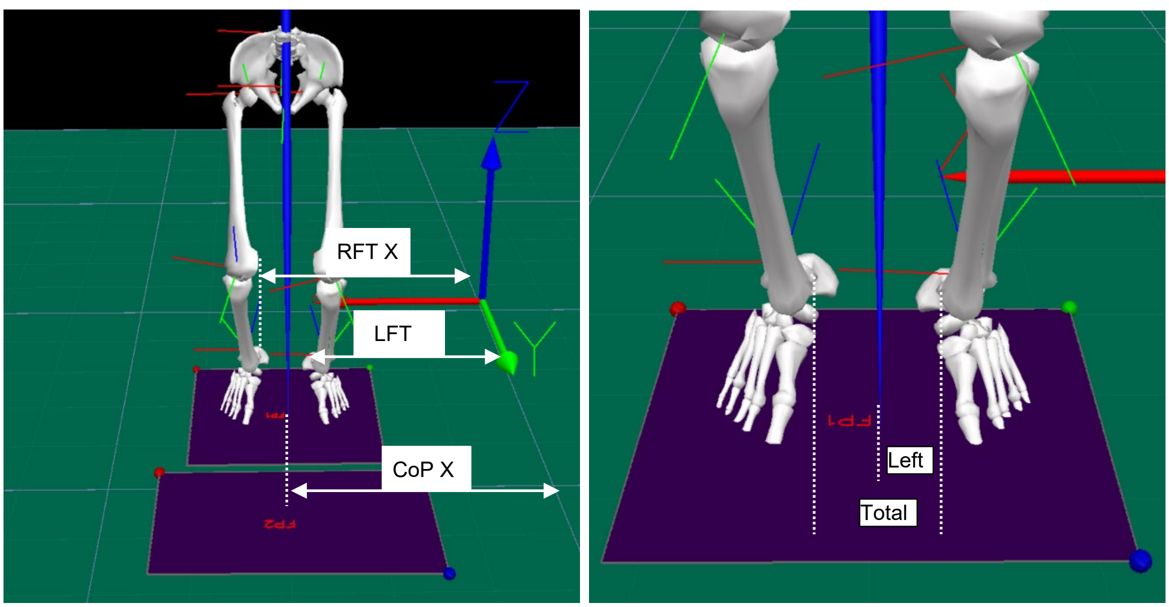


Figure G1 – Calculation of CoP symmetry. CoP location data combined with kinematic data of the foot (left) yields the symmetry data (right).

All participants showed less than 10% deviation from a perfectly centered CoP (50%; **Figure G2)** with an average of 0.494.

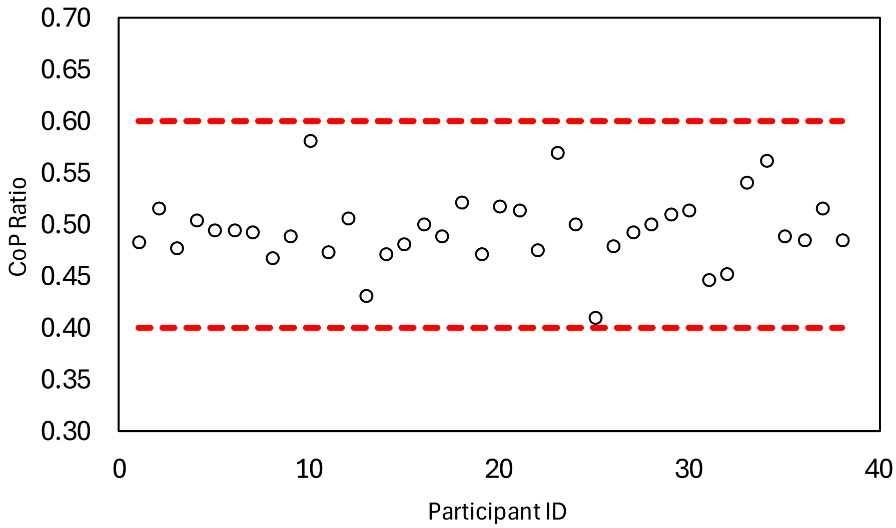


Figure G2 – Ratio demonstrating CoP symmetry at the instant of peak GRF. No participants showed more than 10% deviation in the symmetry ratio. Perfect symmetry, which means the CoP is centered between both feet is represented by a ratio of 0.50.

GRF Symmetry

A similar analysis was conducted with left and right GRFs in the 24 participants (039 onward) who used the stair apparatus with two embedded force plates. As well, seven additional data points are provided from participants who returned after 6 months (**Figure G3**). On average, participants showed 3% lower GRF on the left. This result suggests that it is likely acceptable to use a single force plate to measure inverse dynamics in symmetrical sagittal plane movements.

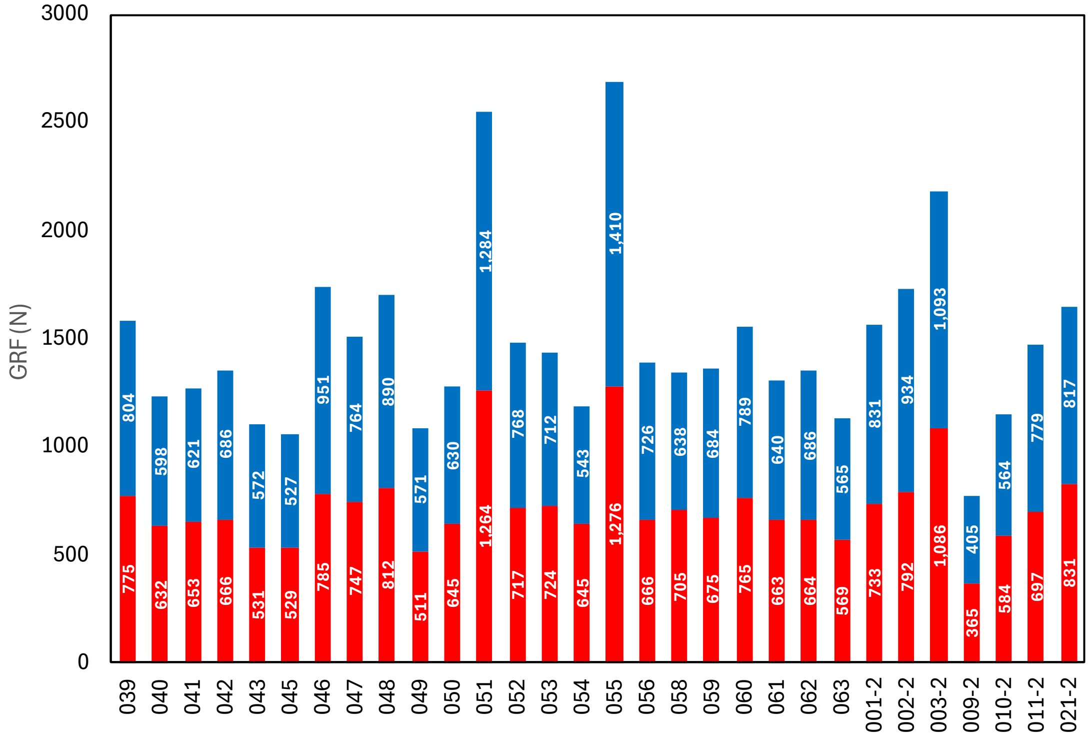


Figure G3 – GRF symmetry across participants with two embedded force plates. Participants demonstrated a 3% lower GRF on the left during CMJ. Participant codes with '-2' indicate return visits.