The Role of Low Back Capacities on Loaded and Unloaded Functional Movements: Squat and Lunge

by

Elizabeth Zafar

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Author's declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Abstract

Variability of movement patterns across individuals have been well documented in healthy young adults. Large heterogeneity of movement patterns within seemingly homogenous populations suggests the possible presence of sub-grouping of individuals. This variability makes it difficult to study and draw conclusions based on group effects, since group means may not be representative of individuals within the group, and especially when subgroups respond differently to interventions. It is also well established that certain movements and movement characteristics are relevant to movement efficiency, tissue exposure, and injury risk, however, it is not fully understood why individuals utilize certain movement patterns over others. It is plausible that physical capacity related differences between subgroups of individuals can help explain differences in movement. As such, this thesis aimed to cluster individuals according to their lumbar movement profiles during functional movements, and then relate characteristic profiles of each subgroup to the low back capacities of strength, muscular endurance, proprioception, and motor control. Additionally, this thesis investigated the effects of introducing a moderate challenge (i.e., loading) to the lumbar movement profiles during functional movements.

Thirty-two healthy young adults (16 M, 16 F) performed two sets of ten repetitions each of squat (SQT) and lunge (LNG) in both unloaded (UL) and loaded (LD) conditions. Additionally, lumbar capacity tests of strength (S), endurance (E), joint position accuracy (P-A), joint position sensitivity (P-S), and motor control (MC) were assessed. State spaces of lumbar angle dynamics for each condition of movement were constructed, then discretized into 48 bins and averaged across repetitions. State spaces were then analyzed using spectral clustering with the number of sub-groups selected based on the strongest silhouette score. Analyses of variance (ANOVAs) testing the effect of sex and group on each capacity test's scores were conducted.

The results of the clustering produced two groups with weak clustering strength in each condition. In both the SQT UL and SQT LD conditions, a significant interaction between sex and group in P-S ($p = 0.01$), and a significant effect of sex in E ($p = 0.04$)

were found. In the LNG UL condition, a significant interaction between sex and group in P-A ($p = 0.04$), and a significant effect of sex in E ($p = 0.04$) were found. Significant interactions between group and sex were found in both P-S ($p = 0.04$) and MC ($p = 0.03$) for the LNG LD condition. Differences in lumbar capacities between groups were related to features of the state spaces, including shape, diffuseness, and intensity of attractors.

This thesis highlighted the importance of physical capacities on movement patterns and affirmed the necessity of characterizing subgroups of individuals within a heterogeneous sample population. This thesis provides a framework for more comprehensive investigations into the relationships between specific capacities and movement profiles.

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Chapter I: Introduction

Movement strategies and coordination patterns contain important biomechanical features that provide information about an individual's tissue exposures, risk of injury, and physical condition. Kinematic and kinetic information have been used extensively to objectively characterize movement differences in various populations, whether for distinguishing dysfunctional movement patterns from healthy movement patterns [Cholewicki et al., 2002, Cook et al., 2006a, Cook et al., 2006b], determining efficient or ideal strategies for performance [Maddox et al., 2020] or prevention of injury [Marras, 2000], or to glean knowledge regarding the mechanisms [Chang et al., 2008] and control [Larson and Brown, 2022] behind the movement. Inevitably, some amount of inter-individual variability is present and expected when conducting biomechanical analyses, not least because there are many factors that interact and adapt within biological systems and there is some level of distinguishing uniqueness about an individual's movement patterns. Indeed, it is not uncommon for people to recognize a close friend from afar, based only on unique anthropometric and movement characteristics. It is therefore not surprising that variability exists within a population. However, when groups of individuals exhibit characteristics of movement that are similar to each other but distinct from other individuals, there is reason to investigate what factors beyond the normal variability within a homogenous population may be the underlying cause of the movement patterns, and whether there are associated implications attributable to these differences. Several studies regarding functional movement, quality of movement, and training effects have discovered large heterogeneity within healthy adult sample populations [Beach et al., 2019, Dufek et al., 1995, Frost et al., 2015a]. When subgrouping of individuals is not accounted for, it is possible that group means do not properly represent individuals within the group [Dufek et al., 1995]. Additionally, it is possible that effects could cancel out or increase variability, resulting in acceptance of a null hypothesis despite the existence of individual effects. Overall, conclusions and future recommendations are difficult to make when variability between individuals is large.

While differences in movement strategy between varying populations (e.g., healthy

vs dysfunction, age populations, etc.) can be expected, why some individuals within seemingly homogenous populations (e.g., age and health status) respond differently to an exposure compared to others has yet to be explained. It is possible that different physical capacities related to movement may render certain interventions effective for some and not for others. Elements of movement such as muscular strength, muscular endurance, proprioception and motor control all interact to promote healthy movement. It is reasonable to hypothesize that for each element there is a range of capacities and that the combination of capacity within each of the elements could be an important factor in explaining the heterogeneous performance in what were previously considered homogeneous samples. As such, in varying environments and task conditions where certain elements are more challenged than others, individuals with higher capacity in those elements would perform differently compared to individuals with lower respective elements.

The overarching goal of this thesis was to work towards elucidating the fundamental reasons for the variety of movement strategies seen between individuals in similar tasks. This thesis aimed to capture the individual low back capacities of muscular strength and endurance, proprioception, and motor control to determine if capacities can explain differences in movement strategy. It was hypothesized that clusters of distinct movement profiles exist within a healthy, young adult population, and that differences in characteristic movement profiles in the lumbopelvic region are related to these low back capacities. An additional aim of this thesis was to explore the use of common movement strategies across different tasks to identify meta-clusters of individuals for more comprehensive characterization of movement traits within individuals. Findings from this thesis help inform individualized testing and programming for young adults and provide insight for future research regarding injury-reducing clinical and workplace interventions for potentially problematic movement strategies.

Research questions & hypotheses

Research question & hypothesis 1

Do healthy young adults with different functional movement profiles in the bodyweight squat (SQT) and lunge (LNG) movements have different scores on low back strength, endurance, motor control, and proprioception tests?

It was hypothesized that individuals in different bodyweight functional movement profile groups will have different scores of motor control and proprioception, but no difference in scores for strength and endurance for each of the SQT and LNG movements. Further, it was hypothesized that no sex differences in capacity would be found between groups.

Research question & hypothesis 2

Does loading of the functional movements to increase the task challenge change the observed (RQ1) relationship between movement profile and low back capacity test scores?

It was hypothesized that individuals in different loaded functional movement profile groups will have different scores of all capacities (strength, endurance, motor control, and proprioception) for each of the SQT and LNG movements. Further, it was hypothesized that sex differences of strength and endurance would be found between groups. As such, movement profiles were expected to be clustered mainly by sex in loaded conditions.

Chapter II: Literature review

Low back function

Muscular strength

Muscles are the main actuators of human movement; they consist of bundles of fascicles, made up of bundles of individual muscle cells called muscle fibers. The contractile components within a muscle cell are called myofibrils and contract in a way that is consistent with the sliding filament, crossbridge cycle and excitation-contraction coupling theories. The sliding filament theory [Huxley, 1969] describes how repeated contractile units called sarcomeres are arranged in series within a myofibril, and contract when the thick myosin protein filaments "slide" by way of the crossbridge cycle and excitation-contraction coupling [Calderón et al., 2014, Sandow, 1952] along the thin actin filaments. With the summation of force produced through this mechanism, a whole muscle is able to produce enough force to move limbs; certainly, muscles throughout the body act in concert to produce functional movements and allow us to complete a variety of tasks.

Various factors are thought to influence the amount of force a muscle can produce. Larger muscle physiological cross-sectional area (PCSA) has shown to be correlated to force production [Fukunaga et al., 2001], though some inconsistencies in the literature suggest that several other factors may also influence force production, including training status [Castro et al., 1995, Sale et al., 1992], contraction type [Herzog, 2014] and morphological characteristics such as muscle fiber type, fascicle length, and pennation angle [Brechue and Abe, 2002, Dons et al., 1979, Fukunaga et al., 2001]. Muscle strength is also dependent on the moment arm of muscles, which is dependent on posture [Tveit et al., 1994]. Age has also been associated with overall isometric muscle strength, with a general decline in strength after 30 years of age [Asmussen and Heebøll-Nielsen, 1962].

Muscular strength tests aim to measure the maximum amount of force able to be acted on in the external environment that is voluntarily produced [Bohannon, 2019].

Muscular strength of the low back can be tested using an isokinetic dynamometer, handheld dynamometer or a loading cell. Isokinetic dynamometry is the gold standard for measuring trunk strength [Delitto et al., 1991, Karataş et al., 2002, Neblett et al., 2003] although it is expensive and thus less practical for common clinical settings [Juan-Recio et al., 2018]. Hand held dynamometry was found to have moderate quality evidence for intra-rater reliability [Althobaiti et al., 2022], though examiner strength when using a hand-held dynamometer has also shown to influence the strength measures, in particular, for large muscle groups that can produce above 120 N of force at the measurement location [Wikholm and Bohannon, 1991]. In a systematic review [Althobaiti et al., 2022], only 4 studies were found that used a digital loading cell to measure trunk extension strength [Essendrop et al., 2001, Mesquita et al., 2019, Paalanne et al., 2009, Pitcher et al., 2008]. None of these studies measured construct validity of their methods, though evidence for intra-rater reliability was deemed to be of high-quality [Althobaiti et al., 2022]. One of the studies [Essendrop et al., 2001] highlighted that between sessions, confidence intervals of isometric tests on a group level were sensitive, while on an individual level they were much less sensitive (requiring about 30% improvement for a smallest detectable difference compared to the 10% required at the group level). These supported results found by [Dieen and Heijblom, 1996], who also demonstrated that changes should necessarily be large (in general, more than 20%) for reliable detection. As such, researchers investigating individual changes in low back strength over time should be aware of the large effect necessary to attribute a change in strength.

Muscular endurance

Muscular endurance (different from the endurance required by long-distance running or other endurance sports which consider the capacity of the cardiorespiratory system) is clinically measured by the length of time in which a person can hold a specific posture, or the number of repetitions a person can complete of a specific dynamic exercise, and it represents the resistance of the muscular system to fatigue. Muscular endurance can

be influenced by both muscle physiological factors as well as by neural factors, including higher level cognitive aspects such as motivation or understanding and familiarity of the task [Jørgensen and Nicolaisen, 1986, Martínez-Romero et al., 2020].

Different muscle fiber types are known to have varying rates of force development, and varying levels of resistance to fatigue. While human muscle fibers lay on a spectrum and have the ability for plastic adaptation, they are generally classified into categories of Type I (slow oxidative), Type IIA (fast oxidative glycolytic), and Type IIX (fast glycolytic) based on their physiological and metabolic function, protein morphology, and velocity of contraction [Pette and Staron, 2000, Plotkin et al., 2021, Schiaffino and Reggiani, 2011]. The combination of fiber types within a muscle will determine the capacity and qualities of that muscle. For example, postural muscles have a higher composition of Type I (slow oxidative) fiber types compared to non-postural muscles, and as such, are better suited to holding contractions over long time periods [Arbanas et al., 2009, Mannion et al., 1997]. In general, males have been found to have greater fatigability of trunk muscles than females as measured by performance outcome and electromyography (EMG)-related fatigue indices, which may be because of differences in fibre type proportions between males and females [Mannion et al., 1997].

Numerous tests of lumbar muscle endurance have been utilized, including the Biering-Sørensen test (BST), upright pulling timed test, semi-crouched lifting test, isokinetic dynamometry, flexion-rotation trunk test, prone isometric chest raise, prone double straight leg raise, and dynamic extensor endurance test [Silva et al., 2005, Juan-Recio et al., 2018, Martínez-Romero et al., 2020]. Of all these tests, two emerge from the literature as the most practical for use: the Biering-Sørensen test, and the upright pulling test [Silva et al., 2005, Jørgensen and Nicolaisen, 1986, Martínez-Romero et al., 2020]. The BST [Biering-Sørensen, 1984], is most commonly used since it is easy to implement and doesn't require special training or equipment. It involves lying prone on a bench with the anterior superior iliac spine (ASIS) lined up with the edge such that the upper body is unsupported and securing the lower body to the bench with inelastic straps. The outcome measure is the number of seconds, up to 240 seconds, the individual is

able to hold the upper body horizontally. A modified version of the BST removes the upper limit of 240 seconds [Jørgensen and Nicolaisen, 1986]. Importantly, low scores on the BST have been found to be correlated with incidences of lower back pain [Luoto et al., 1995], making it a useful tool for predicting who is at high risk of developing LBP and who may benefit from intervention. On investigation of the BST using EMG however, it was discovered that not only lumbar paraspinal muscles, but also gluteal and hamstring muscles are involved in maintaining the horizontal position especially towards the end of the test when lumbar fatigue is experienced [Clark et al., 2003].

In contrast, the upright pulling test (UPT) [Jørgensen and Nicolaisen, 1986], while similar to the BST in that it assesses endurance of the low back, is completed in an upright standing position with the pelvis supported posteriorly. The individual is instructed to extend their back against an inelastic cable attached on one end to a harness secured around the shoulders and on the other end to a force transducer. The UPT was shown to assess the lumbar spine muscles in more isolation compared to the BST [Clark et al., 2003] and to have higher reliability of electromyography (EMG)-derived fatigue indices [Koumantakis et al., 2001] though reliability of test scores was good (reliability coefficient of 0.82) but not higher than the BST (0.89) [Jørgensen and Nicolaisen, 1986]. The UPT additionally has the benefit of knowing and being able to control for the force of pull since it is completed at 60% of the MVIC force, unlike the BST which is tested using the upper body mass as resistance. However, since the UPT is in an upright posture, it presents a greater challenge to postural control than the BST and requires sufficient stabilization of the pelvis to avoid lower limb involvement. Comparatively, the UPT results in fatigue in about a third of the time of that resulting from the BST [Jørgensen and Nicolaisen, 1986].

Proprioception

Proprioception is the sense of one's relative body positioning and movement. Various receptors including Golgi tendon organs [Roberts et al., 1995], muscle spindles [Buxton and Peck, 1989], and mechanoreceptors [Jiang et al., 1995, Roberts et al., 1995]

play a role in providing feedback to the central nervous system (CNS) regarding the position and/or movement of the body. This is crucial not only for modulating movement behavior (i.e., proprioceptive neurons modulate interneurons and motor neurons in the spinal cord) [Jankowska, 1992] and proper execution of planned tasks [Buxton and Peck, 1989, Hillier et al., 2015, Peter Reeves et al., 2007], but also for preventing injury related to hypermobility or instability of joints [Jiang et al., 1995, Riemann and Lephart, 2002]. Various tissues in the lumbar region have been found to contain proprioceptive receptors, including ligaments [Yahia et al., 1988], thoracolumbar fascia [Mense, 2019, Yahia et al., 1992], intervertebral discs and endplates [Fields et al., 2014, Yamashita et al., 1993], and paraspinal [Ford et al., 1988] and intrinsic postvertebral [Amonoo-Kuofi, 1982] muscles. Interestingly, it was found that two different tests of upper extremity proprioception (movement discrimination threshold test and arm position matching test) were uncorrelated [Lowrey et al., 2020] suggesting that perhaps proprioception has multiple aspects and requires a suite of tests to capture the full proprioceptive function rather than a single test.

Spinal proprioception has been measured using passive (PJRE) or active (AJRE) joint repositioning error and threshold to detection of passive motion (TTDPM) [Ghamkhar and Kahlaee, 2019, Lee et al., 2010] and has been of interest in the clinical population since low back pain patients have been found to have poorer repositioning accuracy [Georgy, 2011], higher TTDPM [Lee et al., 2010], and altered proprioception-related postural control strategies [Claeys et al., 2011], though in a meta-analysis, pain and disability were not correlated to proprioception measured by active joint repositioning, TTDPM, or passive joint repositioning except for only to a small degree correlated to the absolute error in passive lumbar flexion and extension repositioning [Lin et al., 2019]. However, another meta-analysis found that the TTDPM was worse, and the joint reposition sense poorer in patients compared to controls when tested actively in a seated position, but not actively in a standing position or passively in a seated position [Tong et al., 2017]. Additionally, active and passive trunk repositioning error at 30 degrees of flexion, and passive repositioning at 60 degrees of flexion were poorer between

athletes with patellofemoral pain and healthy athletes [Armaki et al., 2021], suggesting potential distally directed effects with poorer spinal proprioception. In healthy individuals, spinal position sense measurements were reproducible in upright and flexed postures to within a mean of 3.79±2.56 and 5.27±3.47 degrees respectively, in the sagittal plane [Swinkels and Dolan, 1998].

Motor control

Motor control (also neuromuscular control) is a term that, in the area of human biology, broadly refers to the process of planning and execution of movement and the underlying control processes. The central nervous system (CNS) is understood to be the center where this control happens: movement is planned and motor commands are initiated, modulated based on sensory integration and updating of conditions in the external environment, and relayed to motor neurons in the peripheral nervous system (PNS). Motor neurons, together with the muscle fibers they innervate, are called the motor unit and follow an "all-or-none" principle where, in healthy individuals, if the neuron threshold is met, all of the associated muscle fibers will contract maximally (Lucas, 1909). Force production in the whole muscle depends on the number and type of motor units recruited. Based on Heneman's size principle [Enoka and Stuart, 1984, Henneman and Olson, 1965], in general motor units with the lowest activation threshold are recruited first (typically containing type I fibers), followed by medium and higher-level activation thresholds (type IIA and IIX fibers).

Changes in motor units occur throughout the lifespan, typically with fiber denervation occurring with age and associated with atrophy and loss of muscle fibers [Wilkinson et al., 2018]. However, compensatory changes can occur after injury or other denervation mechanisms such as inactivity or aging, with modulation of motor units due to changed neuromuscular activation [Gordon and Pattullo, 1993] and axonal sprouting occurring in partially denervated motor units [Jones et al., 2022, Siu and Gordon, 2003].

Neuromuscular control has been linked to injury risk of the lower limb, likely due

to the need to meet demands of rapidly changing trunk direction in cutting, stopping and landing movements [Ste Croix et al., 2018, Zazulak et al., 2007]. Training, on the other hand, has shown to improve neuromuscular control, improving known injury risk factors in the knee [Ste Croix et al., 2018, Hewett et al., 2017], with particular effectiveness when implemented during adolescence [Williams et al., 2021].

No single assessment of motor control exists that encompasses all its aspects. As such, studies often use multiple tests to compare differences in whole-body neuromuscular control [Read et al., 2016, Williams et al., 2021]. For example, one study included the Y-balance test, single-legged hop jumps, drop jump, 10- and 20-meter sprints, agility test and range of motion testing to compare between neuromuscular training protocols for youth football players [Ayala et al., 2017]. Other common field tests for motor control include the Functional Movement Screen (™) [Baeza et al., 2017, Cook et al., 2006a, Cook et al., 2006b], Tuck Jump Assessment [Sonesson et al., 2022], time to stabilization [Zech et al., 2014] and postural sway [Zech et al., 2014]. Neuromuscular control specifically of the trunk has been investigated by various methods, including a seated stabilizing task [Cholewicki et al., 2000, Reeves et al., 2020], muscular response latency to a trunk unloading perturbation [Cholewicki et al., 2005], muscle recruitment patterns [Dieën et al., 2003], trunk position and force tracking and stabilization tasks [Reeves et al., 2014, Reeves et al., 2018], and motion smoothness [Miller et al., 2023, Perrott et al., 2021].

Movement strategy variability

A notable amount of inter-individual variability in movement behavior has been demonstrated in numerous studies, particularly in functional and whole-body movements [Beach et al., 2019, Frost et al., 2015a, Morriss et al., 1997, Rodano and Squadrone, 2002], but also in more isolated movements such as lumbar flexion and extension [Larson et al., 2018]. Moreover, it is often found that not only highly variable, but distinct and differing responses between individuals result from the same exposure or task [Dicesare et al., 2020, Frost et al., 2015b, Larson et al., 2018]. This potentially leads

to false insignificant findings in various interventions and experiments using population means, or difficulty interpreting significant results and making general conclusions about groups.

Movement subgroups and low back pain

Four main subgroups of movement patterns in a sample population were discovered from a standing flexion and a seated task [Laird et al., 2018]. Their sample included people with and without a history of low back pain, but rather than looking at differences in movement variables and coordination between healthy and low back pain patients, they formed groups first based on the movement profiles using an unsupervised machine learning algorithm and then compared between the resulting groups for low back pain and movement variables. The analysis resulted in the formation of four subgroups which they labeled: 1) standard, 2) lumbar dominant, 3) pelvic dominant, and 4) guarded. While some subgroups contained a higher percentage of low back pain patients compared to other subgroups (in particular, the guarded subgroup were all low back pain patients), all groups included at least 26% low back pain patients, and most (three of four) groups included at least 17% back healthy participants. Nevertheless, patterns represented by the subgroups with increasing levels of LBP patients were not surprising, considering that low back pain patients have also previously been found to have a smaller range of lumbar spine motion [Laird et al., 2019, Laird et al., 2014, Shum et al., 2005], slower movement time [Laird et al., 2014], and an altered flexion-relaxation phenomenon in the low back extensor musculature [Colloca and Hinrichs, 2005, Ippersiel et al., 2021, Neblett et al., 2003, Shirado et al., 1995]. Additionally, the dispersion of healthy and LBP patients into the standard, lumbar dominant and the pelvic dominant subgroups may explain the range of results in lumbopelvic coordination that have been found in the literature regarding its difference between healthy and LBP patients [Ippersiel et al., 2021, Laird et al., 2019, Mokhtarinia et al., 2016, Sánchez-Zuriaga et al., 2015, Shojaei et al., 2020]. These results suggest that subgroups of movement-related patterns may be found within a sample population and

that movement patterns may relate to low back pain but cannot be fully explained by the presence or absence of low back pain. Additionally, they show that a group of back-healthy individuals can also present with varying movement characteristics which may have subgrouping tendencies. In a systematic review, it was found that low back pain individuals tended to show higher movement variability than control groups [Laird et al., 2014]. However, high variability between studies, small between-group differences, and high within-group differences cast some uncertainty on using only movement assessment as evidence of dysfunction at the individual level [Laird et al., 2014]. In one study [Ippersiel et al., 2021], altered lumbo-pelvic coordination in low back pain patients was significant only when 3 distinct outliers presenting with a specific active-extension pain pattern were included in the sample; no difference in coordination was found when the outliers were removed from the dataset. These findings further support the ongoing need to properly sub-classify individuals with low back pain, but also indicate that research may benefit additionally from movement profiling and classification of individuals without a history of low back pain [Ippersiel et al., 2021, Laird et al., 2018].

Functional movement tasks

Studies investigating whole-body and functional movements have similarly produced evidence of large inter-participant variability and individual movement strategies [Babic et al., 2016, Dicesare et al., 2020, Frost et al., 2015a, Frost et al., 2015b]. External factors such as load, speed [Flanagan and Salem, 2008, Frost et al., 2015b], and task [Granata et al., 1999, Norasi et al., 2019] as well as internal factors such as attention [Katsuhira et al., 2013, Peh et al., 2011], task perception [Dufek et al., 1995, Frost et al., 2015a], experience [Granata et al., 1999], and physical characteristics [Clark et al., 2016] have been found to modulate an individual's movement behavior. However, the way some of these factors modulate movement behavior is not consistent between individuals across tasks or within tasks for different conditions [Frost et al., 2015b]. As such, conclusions based on group data regarding specific demands of tasks, adapta-

tion strategies that are necessary when tasks are challenged, and further recommendations for reducing injury risk become difficult to make when such variable responses are seen between individuals [Beach et al., 2019, Dufek et al., 1995]. For example, sagittal plane spine motion in a squat task showed a meaningful change in 40% of participants, though half of them had a positive change while the other half had a negative change [Frost et al., 2015b]. Additionally, another study conducted by the same researchers found that while focused-movement training was more effective than general fitness-focused training in improving movement patterns, not all individuals responded similarly to the intervention, with almost half as many individuals within the group producing negatively associated changes in movement as individuals producing positively associated changes [Frost et al., 2015a].

These results are likely to be related to both individual physical capacities as well as the demand of specific tasks; that is, the way these factors interact may be producing such varied results. However, how individuals behave during functional and workplace tasks was still shown, in general, to be trainable and induce positive effects when training was focused on movement quality [Frost et al., 2015a]. Indeed, it is likely that adjusting training protocols to individualize based on the specific presentation of movement strategies prior to the training could produce more consistent positive results in movement qualities. Furthermore, improving the quality of functional movement (e.g., movement strategy), is an important factor in reducing the risk of potential injury, since cumulative damage can occur due to poor strategies, causing increases in the risk of overuse injury [Callaghan and McGill, 2001, Hewett et al., 2005].

Certainly, the need for the individualization of assessment, injury risk management and training is evident. There is some evidence that strategies in net joint moment contributions in the lower limb during a barbell back squat converge at high load magnitudes with smaller variance between individuals [Flanagan and Salem, 2008]. Overall, the variability in this study was quite large and no subgroups of participants were detected based on movement strategy [Flanagan and Salem, 2008]. Considering the body of evidence in the literature regarding inter-participant movement variability,

it is possible that the barbell back squat exercise task constrains movement in a way that requires a specific movement strategy, regardless of potential differences and sub-groupings of preferred movement patterns between individuals completing other tasks. Additionally, it is plausible that participants all completed the barbell back squat using a strategy that was similarly coached and practiced in an exercise or workout environment based on principles of safe movement, especially considering that participants all had at least a year of experience back squatting [Flanagan and Salem, 2008], and individuals can modulate their movement based on knowledge of intended criteria [Frost et al., 2015a]. Indeed, improvement of posture [Scannell and Mcgill, 2003] and movement variables (e.g., reduced frontal plane knee motion and spine flexion) [Frost et al., 2015a] were achieved when feedback was given by coaches during a 12-week training program, though again with considerable movement variation in responses found at the individual level. This suggests that a deeper understanding of individual movement strategies can inform the prescription of effective training protocols aimed at reducing high stress or strain loads on joints to reduce the risk of injury.

Movement coordination: Squat and lunge

In the double-legged squat exercise, relative phase measures in different phases (lifting and lowering) of the exercise were found to be affected differently by load, with lumbar spine extension lagging further behind knee extension in the lifting phase as load increased, while in the lowering phase lumbar spine flexion led knee flexion with no effect of load [Scholz and McMillan, 1995]. The authors found a similar pattern between the hip and knee joint relative phases, though increased lag effect of load on the lifting phase was smaller [Scholz and McMillan, 1995]. Interestingly, the authors found that individual differences in lift strategy were found within their population such that two general groups seemed to form, with 2 additional outliers; as such, a secondary analysis was conducted investigating these differences on the two main groups [Scholz and McMillan, 1995]. Group differences in knee-lumbar spine relative phase were found to be greatest at early stages of the lift cycle, particularly at lighter loads,

and disappeared later in the lift cycle. Specifically, they found a greater lag of lumbar spine extension to knee extension in one group compared to the other. Not surprisingly, another study investigating movement strategy of the squat exercise found similar large standard deviations and individual differences within their sample populations, which were found to predominantly involve couplings with the trunk [King and Hannan, 2019]. In another study, a pattern recognition and machine learning method was used to find different movement strategies [Remedios et al., 2020]. It was found that applying a Gaussian mixture model to the results of reduced-dimension 3D kinematic data (through principal component analysis) produced three phenotypes for the squat [Remedios et al., 2020]. Typically scored kinematic features were significantly different between groups, but none were able to fully explain the group differences.

The lunge movement is one of the functional movements that make up the widespread Functional Movement Screen ™ [Cook et al., 2006a] and has since been recognized as an important and reliable functional movement that can be used as an assessment tool, in particular, to analyze biomechanical variables such as ground reaction force, knee joint, angle, moment, and power [Alkjær et al., 2009], as well as impulse, contact time, and stride distance [Luque-Siles et al., 2016]. However, when compared to other tests of balance (center of pressure), speed (sprint), and power (maximum vertical jump), no significant correlation was found with any test [Hartigan et al., 2014]. It is possible that important distinctions regarding balance, power, and speed as challenged in the lunge movement require detailed observation that is more possible with the very specific functional movement task compared to alternative field tests in which compensations throughout the whole body may be able to affect performance outcome measures.

No previous literature was found regarding the lumbopelvic intersegmental coordination patterns during a lunge movement. Lower limb coordination, however, was found to have more variation in coordination pattern in the forward lunge when compared to a deadlift and step-up movements which presented with substantially in-phase movement [Romanazzi et al., 2015]. Specifically, the forward lunge showed $\pm 30^{\circ}$ (in-phase) continuous relative phase (CRP) angle for 11%, 10%, and 16% of the movement

duration for the knee/hip, ankle/knee, and ankle/hip couplings, respectively and had two distinct movement periods of mostly out-of-phase movement associated with the swinging phases of the lunge [Romanazzi et al., 2015].

In summary, an overwhelming body of literature points to the presence of movement strategy variability between individuals (especially during whole-body, functional movement tasks), and the tendency for sample populations to form subgroups that behave differently and respond to interventions differently. However, a large part of this inter-individual variability is largely unexplained. Various factors have been proposed to partly explain why differences occur (e.g., load, speed, familiarity of task, physical characteristics, etc.), but they are not satisfactory in explaining the differences. Additionally, interventions targeting specific capacities and movement training have been implemented, though not using a stratified approach that accommodates for subgrouping of individuals. Yet, evidence that physical capacities (e.g., strength, endurance, motor control, and proprioception) are related to movement patterns exists. As such, there is a large gap in the understanding of how individuals cluster with regards to their movement strategy, and the reasons underlying these groupings. Accordingly, it is difficult to implement effective training for individuals for the purpose of improving movement patterns and reducing the risk of injury. Further investigation into both the subgrouping of individuals based on movement pattern as well as profiling of physical capacities associated with these subgroups is warranted.

Chapter III: Methodology

Overview

This study was conducted with a cross-sectional design, whereby low back capacities and functional movement task strategies of each participant were measured on one day within a 3-hour lab session. The research protocol was approved by the University of Waterloo Research Ethics Board under study #45207. Informed consent and clearance for exercise participation and study inclusion was obtained upon immediate arrival of the participants. Anthropometric and other physical characteristics of participants were collected, including sex, age, height, mass, self-reported physical activity level, and leg dominance. Following this, participants were properly instrumented and the appropriate calibration trials completed. Once ready to begin, the study consisted of participants completing two functional movements, squat (SQT) and lunge (LNG), in both loaded (LD) and unloaded (UL) conditions as well as various lumbar capacity tests for: 1) proprioception, 2) motor control, 3) strength, and 4) endurance. Due to efficiency and flow of the protocol, as well as the nature of the tests, the protocol started with proprioception and motor control tests, followed by the UL then the LD conditions of each exercise in a randomized order, and lastly by the strength and endurance tests. All data and analyses were completed using Python 3.11 unless otherwise stated.

Participants

Healthy individuals (17 M, 17 F) aged 18 – 35 years were recruited to participate in the study. Individuals who had prior low back surgery were excluded from the study, as were individuals who have had a lost time work related low back episode or sought medical attention for low back pain in the year prior. A physical activity readiness questionnaire (PAR-Q) and a lower extremity functional scale (LEFS) were administered before the data collection to ensure participants were not placed at greater risk due to participation [Binkley et al., 1999, Warburton et al., 2011]. Exclusion criteria included achieving less than 70/80 on the LEFS or checking YES on any of the PAR-Q

questions, however no participants were excluded on this basis. Two participants (1M, 1F) dropped out of the study due to personal time constraints or forgetting medication at home.

Instrumentation

Motion capture

Kinematic data were captured in three dimensions throughout the lumbar capacity testing and the functional movement protocol using an active optoelectronic motion capture system (Optotrak Certus, Northern Digital Inc., Waterloo ON, Canada). Five camera banks (three cameras each) were used to capture the collection space. During laboratory preparation, cameras were calibrated to the collection space using a rigid cube containing 16 infrared-emitting diodes which was waved around the entirety of the operational volume of interest for 60 seconds. A successful calibration was achieved when a root mean square (RMS) error of less than 0.5 mm was captured by the cameras. A second static calibration was completed to define axes in the global coordinate system, following recommendations by the International Society of Biomechanics (ISB) with +X facing anteriorly, +Y facing superiorly, and +Z facing laterally to the right [Wu and Cavanagh, 1995]. Kinematic data sampling rate was set at 60 Hz.

A marker cluster consisting of at least four markers on a lightweight rigid body was secured to each of the following segments on the participant's body and used to track the segment motion: thorax, lumbar spine, pelvis, and bilateral foot, shank, and thigh. Clusters were secured to the skin using double sided tape (Indoor Carpet Tape, Scotch, St. Paul, MN, USA), medical tape (Hypafix, BSN Medical, Hamburg, Germany), and Velcro straps to avoid movement between clusters and skin. Bony landmarks were palpated and digitized in reference to the cluster locations using a probe containing four markers to define segments, produce a skeletal model of the kinematic data, and allow for calculation of segment positions and joint angles. The location of marker clusters and their associated digitized landmarks are presented in Table 1.

Table 1: Segment marker clusters and associated bony landmarks

Segment	Cluster Location	Digitized Landmarks
Thorax (THX)	7th Thoracic Vertebra	Bilateral acromion processes Bilateral iliac crests
Lumbar Spine (LUM)	1st Lumbar Vertebra	Bilateral 12th ribs Bilateral iliac crests
Pelvis (PEL)	Sacrum	Bilateral iliac crests Bilateral anterior superior iliac spines (ASIS) Bilateral posterior superior iliac spines (PSIS) Bilateral greater trochanter
Right Thigh (RTH)	Right femur	Right greater trochanter Right medial and lateral femoral epicondyles
Light Thigh (LTH)	Left femur	Left greater trochanter Left medial and lateral femoral epicondyles
Right Shank (RSK)	Right fibula	Right medial and lateral tibial condyles Right medial and lateral malleoli
Left Shank (LSK)	Left fibula	Left medial and lateral tibial condyles Left medial and lateral malleoli
Right Foot/Shoe (RFT)	Right anterolateral talus	Right medial and lateral malleoli Right 1st and 5th metatarsal heads Right distal phalanx of 2nd toe Right calcaneal tuberosity
Left Foot/Shoe (LFT)	Left anterolateral talus	Left medial and lateral malleoli Left 1st and 5th metatarsal heads Left distal phalanx of 2nd toe Left calcaneal tuberosity

A sequence of participant calibration trials was completed: 1) an upright standing trial, 2) a maximum standing flexion trial, 3) a maximal standing extension trial, 4) a dynamic functional movement trial for locating the hip joint center [Camomilla et al., 2006, Schwartz and Rozumalski, 2005], and 5) a dynamic functional movement trial for locating the knee joint center [Schwartz and Rozumalski, 2005]. For calibration trial 1), participants were instructed to "stand up straight" and hold as still as they can in that position for a duration of 5 seconds. For trials 2) to 4), participants gave verbal affirmation of maximal range before beginning the 5 second trial. The duration of calibration trials 4) and 5) were 20 seconds each, with instructions to move their limb back and forth throughout the joint range (forward, front diagonal, sideways, back diagonal, backwards, clockwise and counterclockwise circles for hip, and flexion/extension for

knee).

Load cell

A uniaxial linear variable differential transducer (LVDT) was used for the lumbar strength and endurance capacity tests. The LVDT was calibrated manually using a two-point linear calibration method with no weight (resting on a surface with the measuring axis perpendicular to the pull of gravity) and a known weight (103 N). Force measured from the load cell was sampled at 2500 Hz.

Force plates

Three force plates (Advanced Mechanical Technology, Inc, Watertown, MA) were used during the functional exercise trials to measure ground reaction forces (GRF). Calibration was completed using the shunt voltage and force values provided by the manufacturers. Participants completed exercises with each foot on a separate force plate, and GRF was sampled at 300 Hz.

Data collection

Functional movement task protocol

Prior to the functional movement protocol, foot placement and exercise depth targets were determined based on the participants' anthropometry. Stance width was set at shoulder width for SQT, and step length for LNG was set at the length of shank height between the heel of the front foot and the toe of the back foot, as recommended in the Functional Movement Screen (™) for the deep squat and in-line lunge, respectively [Cook et al., 2006a]. The standardization of the LNG was taken from the in-line lunge; however this study used a regular forward lunge. Additionally, for the SQT, participants were first positioned with their feet at shoulder width but were then allowed to adjust their toes according to their preference, while keeping their heels in position to allow for a more natural and comfortable squat for varying hip morphologies. Tape was placed on the ground to mark foot placements. Depth for each exercise was

standardized as well, with the target SQT depth to roughly 85° (just below 90°) of hip flexion angle, and LNG depth to roughly 3 cm off the ground for the back knee. Participants were instructed to assume the set depth position for SQT (isometric hold for a maximum of 5 seconds), and a researcher visually confirmed the depth while a second researcher placed a target on a vertical bar at eye level. For LNG, the participant was asked to kneel on a 3 cm block in the lunge position, and a second target was placed at eye level. The vertical bar was placed in front of the participant during the exercise conditions, and they were instructed to lower until their eyes met the target height. Upper limb positioning was standardized by instructing participants to hold the weights in a goblet position, and a wooden gymnastics ring was supplemented instead of the weights for unloaded conditions.

Participants completed the SQT and LNG exercises in a randomized order, first in UL, then under LD conditions. The load was determined by percentage of bodyweight (BW), with %20 BW for SQT and %15 BW for LNG. For each exercise condition, two sets of 10 repetitions at 0.25 Hz (4 seconds per repetition) were completed with a 1-minute seated break for UL and a 2-minute seated break for LD conditions between sets. A metronome was set at 60 bpm, with the instruction that each rep should take 4 beats, but the movement should be natural (i.e., The movement should not be "choppy" such that each rep is divided into 4 equal parts, which is not natural especially in the LNG exercise). Each exercise was demonstrated visually by a researcher, and given verbal cues that are typical for the respective exercise (e.g., SQT: "keep your knees in line with your feet and lower as if you're sitting back into a chair", LNG: "lower your back knee while keeping your trunk upright"). Asymmetrical exercises were performed on the dominant leg, and participants completed one to two repetitions before the beginning of the first set of each exercise to familiarize themselves with the pace and movement, as well as so that the researcher could correct any misunderstandings on the movement pattern or instructions. After each set, participants were asked to rate the intensity of the whole set based on a rate of perceived exertion (RPE) scale of 1-10.

Low back capacity testing

A proprioception test was completed to allow for a measure of not only joint position accuracy, but also sensitivity to joint position changes. For this test, participants began on all fours (i.e., with hands, knees, and feet touching the ground, and with hands aligned under the shoulders and knees aligned under the hips). Participants were allowed some self-discretion with hip width, to allow for comfort and full range of motion. To begin the test, participants completed movement through full lumbar flexion and extension ranges (to allow for "gauging" of the full range), then self-selected what they perceived to be the halfway point. Once verbally confirming their selected posture, a manual square wave trigger was used to mark the timing within the trial, and the participant was then asked to keep their spinal posture stable while sinking their hips back to their heels, effectively moving into what is commonly known as "child's pose". Once the participant felt a change in their lumbar spine posture, which inevitably happens when moving deeper into the child's pose, the participant verbally confirmed this sensation, and the manual square wave trigger was once again applied. Three trials of the test were completed, each within 35 seconds. From this test, a measure of position accuracy (P-A: calculated as the average difference in lumbar angle between "actual" halfway and "perceived" halfway positions, normalized to the overall range of motion) as well as sensitivity (P-S: calculated from the average difference in lumbar angle between the perceived halfway and the perceived change positions).

The motor control task (MC) consisted of a flexion-extension task which has previously been described [Granata and Gottipati, 2008, Larson et al., 2018]. From this task, a measure of motion smoothness was calculated. Participants stood with their feet shoulder width apart facing two targets, one at the level of the shoulders and the other at the level of the knees (Figure 1). The first target was within arm's reach when the participant arms were outstretched anteriorly, with palms together. The second target was placed 50 cm in front of the knees. A restraint at the level of the anterior superior iliac spine (ASIS) was used to constrain movement of the pelvis. Participants alternately touched the two targets for 25 repetitions in time with a metronome at 60 beats

per minute (bpm), with the participant cycling through one flexion-extension repetition at a rate of 0.25 Hz. This allowed for control of pace and gave the participant an auditory target. Additionally, participants were instructed to try to keep their movement as smooth as possible (i.e., avoid "choppy" movement).

Motion smoothness was calculated using the modified spectral arc length (SPARC) method described by [Balasubramanian et al., 2015]. Smoothness of filtered kinematic angular position data was calculated separately for each segmented flexion movement and extension movement using the following equation for spectral arc length (SAL), defined for a given speed profile $v(t)$ as:

$$SAL = -\int_0^{\omega_c} \left[(\frac{1}{\omega_c})^2 + (\frac{d\hat{V}(\omega)}{d\omega})^2 \right] d\omega \tag{1}$$

$$\hat{V}(\omega) = \frac{V(\omega)}{V(0)} \tag{2}$$

Where $V(\omega)$ is the Fourier magnitude spectrum of $v(t)$, $\hat{V}(\omega)$ is the normalized magnitude spectrum, normalized with respect to the DC magnitude $V(0)$, and ω_c is fixed to be 40π (corresponding to 20 Hz).

In accordance with the SPARC method, ω_c was upper bound by ω_c^{max} and adaptively selected based on a given threshold $\overline{V}$:

$$\omega_c = min \left\{ \omega_c^{max}, min\{\omega, \hat{V}(r) < \overline{V} \forall r > \omega\} \right\} \tag{3}$$

The overall smoothness was then calculated using a weighted average function:

$$A = \frac{\sum_{i=1}^{N} w_i \lambda_i}{\sum_{i=1}^{N} w_i} \tag{4}$$

$$\sum_{i=1}^{N} w_i \neq 0 \tag{5}$$

Where $w_i > 0$ is the weight given to the smoothness of the i-th movement component λ_i.

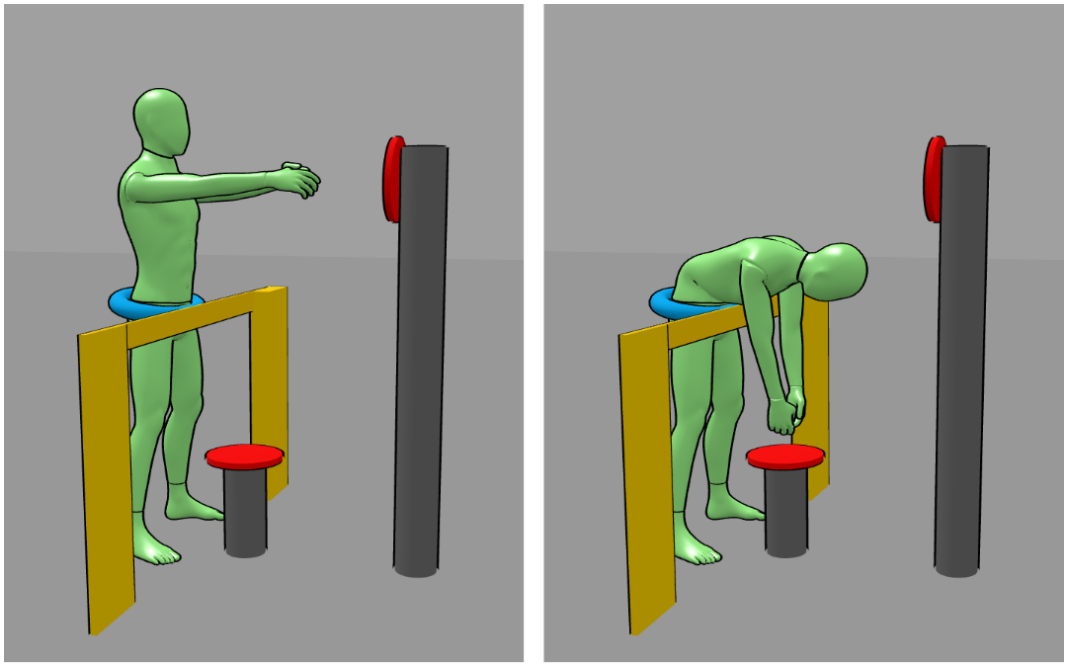


Figure 1: Motor control task (MC) setup. Participants were strapped (blue) to a pelvic support (orange) and alternately pointed towards two targets (red).

For both the strength and endurance tests, participants were positioned in an upright standing position as previously described for low back strength [Biering-Sørensen, 1984] and endurance [Jørgensen and Nicolaisen, 1986] testing. In brief, participants stood with their pelvis supported anteriorly, and with a second inelastic strap around the thorax at the level of T12/L1 connected to a linear variable differential transducer (LVDT) load cell that was fixed to an immovable post anteriorly (Figure 2). To measure lumbar extensor strength capacity, two maximal isometric voluntary contractions (MVICs) of the lumbar extensors were completed with a 2-minute break in between, and the highest output was taken as the individual's maximum strength capacity (S). For each trial, participants were asked to ramp up the force output until they reach their maximum, then to hold for 2 seconds before ramping back down. Force was measured from the LVDT for a duration of ten seconds, within which participants were instructed to complete the entire trial.

The endurance test was completed in the same test position and set up as described for the strength test. To measure the low back endurance capacity, a force value was calculated as 60% of S and served as a target output force to be held until voli-

tional fatigue [Jørgensen and Nicolaisen, 1986]. Participants were instructed to keep the force as close to this target as possible, and a monitor showing the live output of the LVDT and the target force (with lines demarcating ±5% S) were visible to provide feedback. The length of time in seconds, with a maximum of 3 minutes (180 seconds), that the participant was able to hold the isometric contraction above 50% of S was used as the endurance capacity measure (E).

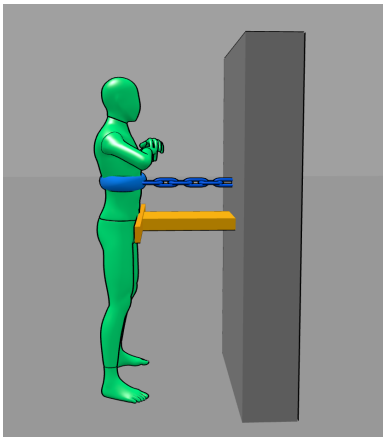


Figure 2: Setup for strength (S) and endurance (E) tests. Participants braced against a pelvic support (orange) and pulled on a load cell via a strap (blue) harnessed at the level of T12/L1.

Data processing

Motion capture kinematic data

Kinematic data (and the associated force plate data) were processed in Visual 3D motion analysis software (Version 6.01.36, C-Motion Inc., Kingston, ON, Canada). Data were first interpolated using a cubic spline interpolation up to a maximum fill of 200 milliseconds [Howarth and Callaghan, 2010]. Next, a 4th order zero-lag Butterworth filter with a low pass cut-off frequency of 10 Hz was applied to both the kinematic and force plate data.

A three-dimensional rigid link model (RLM) was then constructed from the tracked segments and associated bony landmarks (see Table 1). A local coordinate system (LCS) was constructed for each body segment based on ISB recommendations [Wu et al., 2002] and the quiet standing calibration trial was used as reference for the model in

Visual 3D. Functional hip and knee joint centers were defined using the pre-set algorithm in Visual 3D. A Z-X-Y Cardan rotation sequence was used to track segments where flexion-extension movement rotates about the LCS z-axis, abduction-adduction movement rotates about the LCS x-axis, and internal/external rotation about the LCS y-axis. Lumbar and hip joint angles were calculated with respect to the proximal segment.

The low back capacities and associated metrics that were extracted from the kinematic data are presented in Table 2.

Table 2: Extracted lumbar capacity metrics for analyses.

Low back capacity	**Extracted metric**
Proprioception: Accuracy (P-A)	Mean normalized angle error (unitless)
Proprioception: Sensitivity (P-S)	Mean angle error (^{o})
Motor control (MC)	Smoothness of lumbar angle (unitless)
Endurance (E)	Time (s)
Strength (S)	Force (N)

State space construction

Each repetition of the functional movement tasks was analyzed by its state space topology, which is commonly used as a characterization of a system's behavioral trajectory in dynamical systems analyses [Mohamad et al., 2020]. In dynamical systems theory, an area of mathematics that aims to describe complex systems, states spaces are generally used to characterize the current state of the system (in this case, the lumbar angular position and velocity) as well as its evolution function (its trajectory over time). As an example, states tending towards equilibrium are encoded as dense regions in the state space, and commonly referred to as "attractors" of the system. Several studies in biomechanics have applied state space related methods to analyze joint couplings including thorax, lumbar, and hip joints [Maddox et al., 2020, Needham et al., 2014, Zehr et al., 2018].

To characterize the lumbar dynamics during functional movements (SQT, LNG), a state space composed of lumbar angle and lumbar angle velocity was constructed for each movement repetition (Figure 3). Angle values were normalized by maximum

flexion and extension angles within each repetition, and normalized angular velocity values were computed using centered differences of normalized lumbar angles. The state spaces were discretized into 48 bins (normalized width = 0.25) to quantify the frequency and transition of lumbar states. The discretized state spaces were averaged across all repetitions (excluding the first and last repetition in each set) for each movement and condition, resulting in 4 average state spaces per participant.

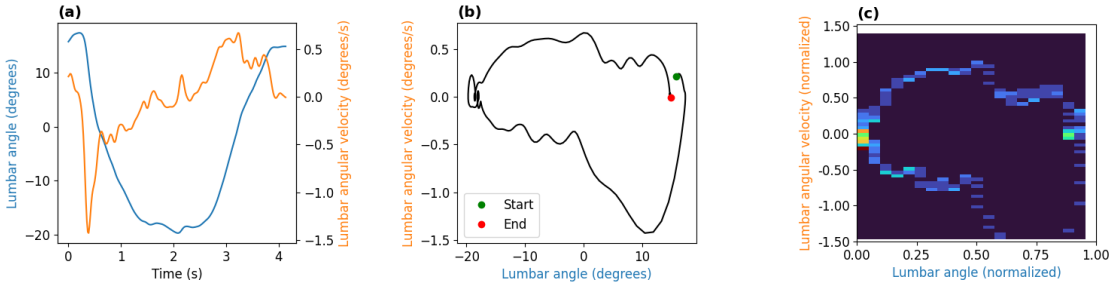


Figure 3: Generation of state space maps from kinematic data: (a) processed kinematic time series data, (b) state space composed of lumbar angle and angular velocity, (c) map of discretized states, colored by frequency of occurrence (high: red, low: blue)

Clustering methods

The characteristic state spaces for each participant, movement, and condition generated a 48 x 1 movement profile vector which was then analyzed using spectral clustering. The similarity between any two movement profiles was quantified by first computing the Euclidean distance between them. So that smaller differences reflected larger similarity, the distances were then passed into a Gaussian radial basis function to produce a final similarity score between 0 and 1. The similarity scores between each pair of participant movement profiles were encoded in a 32 x 32 similarity matrix. The eigenvalues of the similarity matrix were then used to form subgroups of similar movement profiles. The number of groups was then determined by the strongest silhouette score from a range of n=2 to n=10. A meta-clustering of participants was also obtained via a similarity matrix based on the cluster index of individual movements - that is, to group individuals in the same movement clusters across different tasks and conditions.

A silhouette coefficient is a measure computed for each data point (i.e. partici-

pant) and is a value between -1 and 1, reflecting the similarity between the data point with other points in the same cluster. The silhouette score for the overall clustering is taken as the average of the silhouette coefficients for each data point. The strength of the clustering is then evaluated on the range of -1 to 1, with 1 indicating perfect clustering, negative values indicating no valid clustering, and intermediate values indicating corresponding levels of clustering strength as presented in Table 3 [Sasmita et al., 2023].

Table 3: Interpretation of clustering strength from silhouette scores [Sasmita et al., 2023].

Silhouette coefficient	Interpretation
$0.70 < S \leq 1.00$	Strong clustering
$0.50 < S \leq 0.70$	Moderate clustering
$0.25 < S \leq 0.50$	Weak clustering
$-1.00 < S \leq 0.25$	Invalid clustering

Statistical analyses

All statistical tests were completed in R (Version 4.0.3, R Development Team, Vienna, Austria), and an alpha level of 0.05 was used for all analyses.

Data were first tested for normality (Shapiro-Wilk test) and homogeneity of variance (Levene's test). When the data did not meet the assumption of normality, corresponding non-parametric tests were applied (Mann-Whitney U-Test in pairwise comparisons). All data met the assumption of homogeneity of variance.

Correlations between capacity test scores were tested using Spearman's rank correlation to analyze whether individuals ranked similarly across capacities.

For each movement type (SQT, LNG) and capacity (S, E, MC, P-A, P-S), a two-way ANOVA was completed to assess the effects of group (based on the clustering of unloaded movements) and sex on capacity test score. This was then repeated using groups based on the clustering of loaded movements. To analyze the relation of each capacity to meta-clusters of movement groups, a one-way ANOVA was completed to assess the effects of group on capacity test score. One-sample Chi-squared tests were also

run on each meta-cluster to determine the predominance of either sex in each meta-cluster. Pairwise comparisons and Bonferroni corrections were applied post hoc in cases of significance.

Chapter IV: Results

Data were collected from a total of 32 healthy, young adults (16M, 16F). Males demonstrated higher lumbar strength ($p = 0.03, d = 0.69$) and endurance ($p = 0.02, d = 0.80$) scores compared to females, while MC, P-A, and P-S scores were not significantly different between sexes. Ratings of perceived exertion were higher for loaded functional movements compared to bodyweight movements ($p < 0.001, d = 1.53$). A summary of participant physical characteristics and test scores can be found in Table 4.

Table 4: Summary of participant physical characteristics & lumbar capacity scores.

		Age (years)	Height (cm)	Mass (kg)	P-A	P-S (degrees)	MC	E (sec)	S (N)
Males	Mean	21.9	183	78.5	0.209	9.97	3.05	154	769
	SD	3.1	8.2	14.5	0.15	6.02	0.66	42	293
Females	Mean	21.7	168	66.1	0.31	9.3	2.85	109	608
	SD	2.6	9.3	14.8	0.18	8.05	0.42	67	161
p-value		0.9	<0.001	0.02	0.09	0.8	0.3	0.02	0.03
Cohen's d		0.06	1.66	0.84	0.63	0.09	0.36	0.8	0.69

Spearman's rank correlations showed no strong positive correlation between capacity test scores, and medium-low negative correlations between P-A and P-S ($r_s = -0.4, p = 0.02$), and E and S ($r_s = -0.39, p = 0.03$) scores. A summary of correlation coefficients is shown in Figure 4.

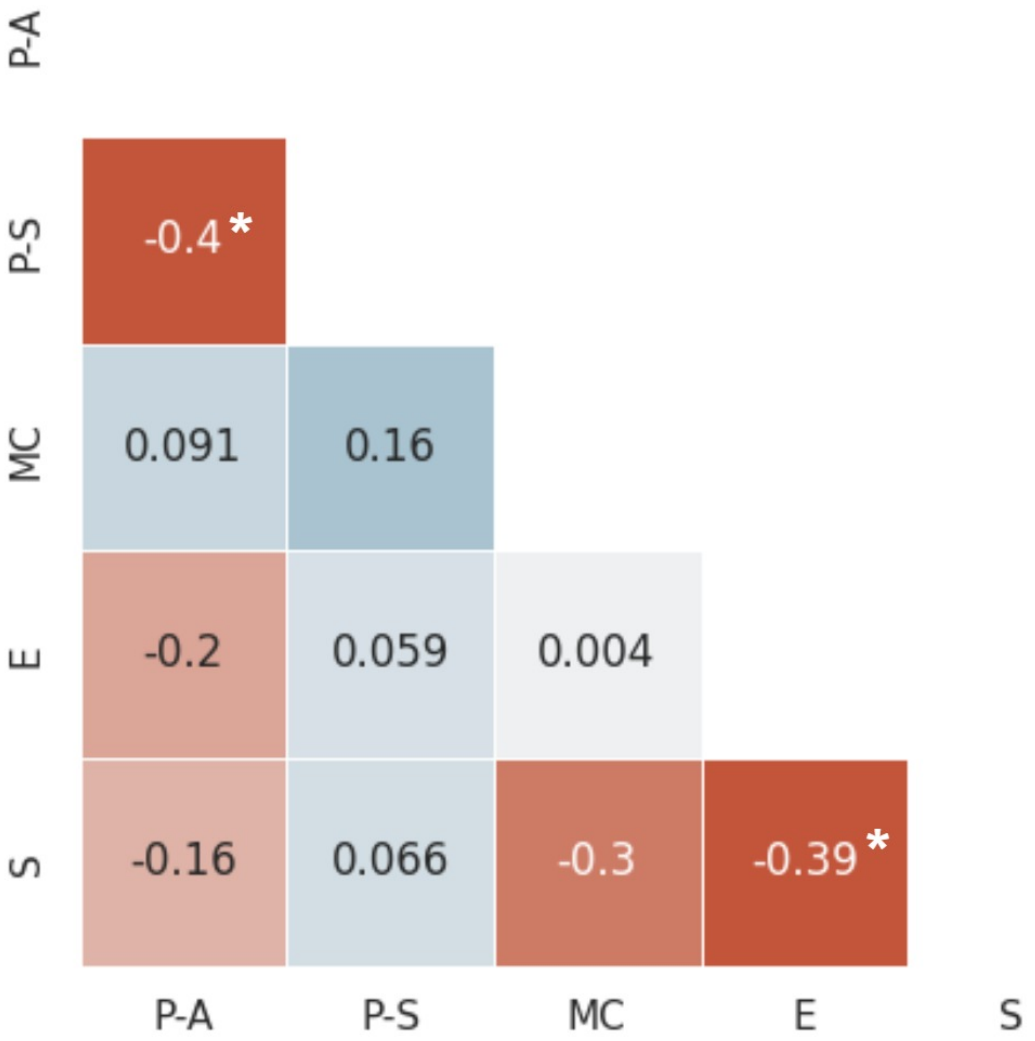


Figure 4: Spearman's rank correlation coefficients between capacity test scores across participants. Significant results ($p < 0.05$) are presented with an asterisk.

Research question 1

Squat

The spectral clustering analysis produced two clusters of participants in the body-weight SQT movement with a silhouette coefficient of 0.29, indicating a weak cluster strength. Group A (n = 14: 6M, 8F) had an average age of 21 ± 2 years, and an average BMI of 22.9 ± 2.3. Group B (n = 18: 10M, 8F) had an average age of 23 ± 2 years, and an average BMI of 23.6 ± 3.9. Average rate of perceived exertion given for UL SQT

movements was 2.2 ± 0.7, which corresponded to the descriptor of "Easy". No differences in age, height, mass, or RPE were found between groups. Average state spaces and extracted lumbar capacity metrics for each group and sex are presented in Figure 5 and Table 5, respectively.

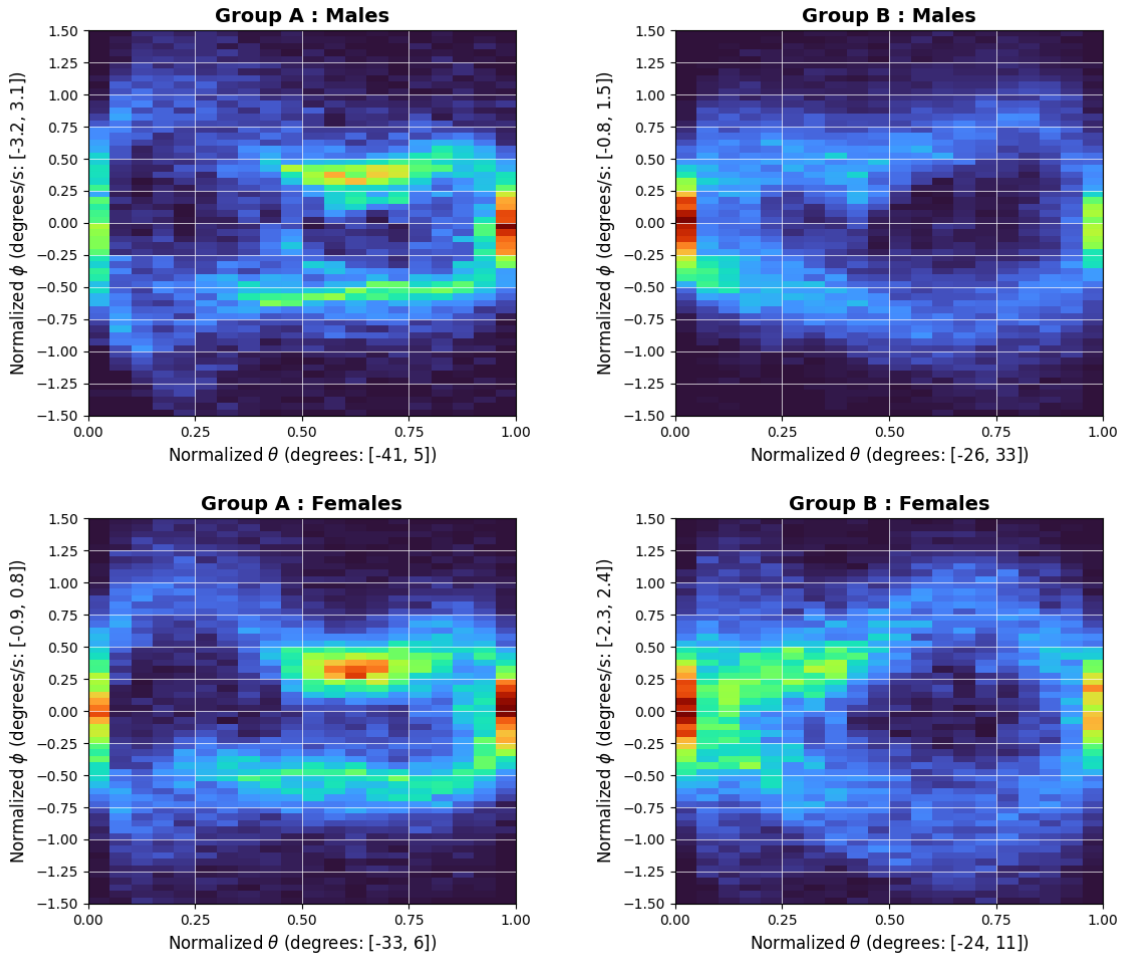


Figure 5: Characteristic state spaces of normalized lumbar angles (θ) and angular velocities (ϕ) for the bodyweight squat, split by group and sex. Color scale indicates low frequency of state occurrence (blue) to high frequency of state occurrence (red).

No significant interaction or main effect of sex or group was found in the SQT UL condition for S, MC, and P-A test scores. However, a main effect of sex was found for E test scores ($F(1, 28) = 4.72, p = 0.04$), though no interaction or main effect of group was found. Pairwise comparisons showed that males had higher endurance times than females ($p = 0.03, r = 0.43$). For P-S test scores, there was no significant main effect of sex or group, however there was a significant interaction between sex and group

($F(1, 28) = 6.82, p = 0.01$). Pairwise comparisons showed that males had higher scores (less sensitivity) than females in group A ($p = 0.04, d = 1.17$), but there was no difference between sexes in group B. Additionally, males in group A had higher scores (less sensitivity) compared to males in group B ($p = 0.02, d = 1.04$). Results from the ANOVAs are presented in Table 6

Table 5: Lumbar capacity profiles for each movement group A and B of the bodyweight squat, split by sex.

Group	Sex	N	P-A	P-S (^{o})	MC	E (s)	S (N)
A	M	6	0.24	14.27	-2.82	162.61	663.15
	F	8	0.39	6.55	-2.91	95.68	574.56
B	M	10	0.19	7.39	-3.18	148.84	832.25
	F	8	0.23	12.05	-2.79	122.63	633.05

Table 6: ANOVA test results comparing lumbar physical capacities across movement groups in the bodyweight squat. Significant results ($p < 0.05$) are presented in bold with an asterisk.

Effect	P-A		P-S		MC		E		S	
	F	p	F	p	F	p	F	p	F	p
Sex	3.98	0.06	0.04	0.84	0.34	0.57	0.13	0.73	1.75	0.19
Group	2.54	0.12	0.97	0.76	0.90	0.35	4.72	**0.04***	3.23	0.08
Sex × Group	1.02	0.32	6.82	**0.01***	1.5	0.23	0.99	0.33	0.43	0.52

Lunge

The spectral clustering analysis produced two clusters of participants in the bodyweight LNG movement with a silhouette coefficient of 0.28, indicating a weak cluster strength. Group A (n = 15: 9M, 6F) had an average age of 22 ± 3 years, and an average BMI of 23.6 ± 3.0. Group B (n = 17: 7M, 10F) had an average age of 22 ± 3 years, and an average BMI of 23.1 ± 3.5. Average rate of perceived exertion given for bodyweight LNG movements was 2.2 ± 0.9, which corresponded to the descriptor of "Easy". No differences in age, height, mass, or RPE were found between groups. Average state spaces and extracted lumbar capacity metrics for each group and sex are presented in Figure 6 and Table 7, respectively.

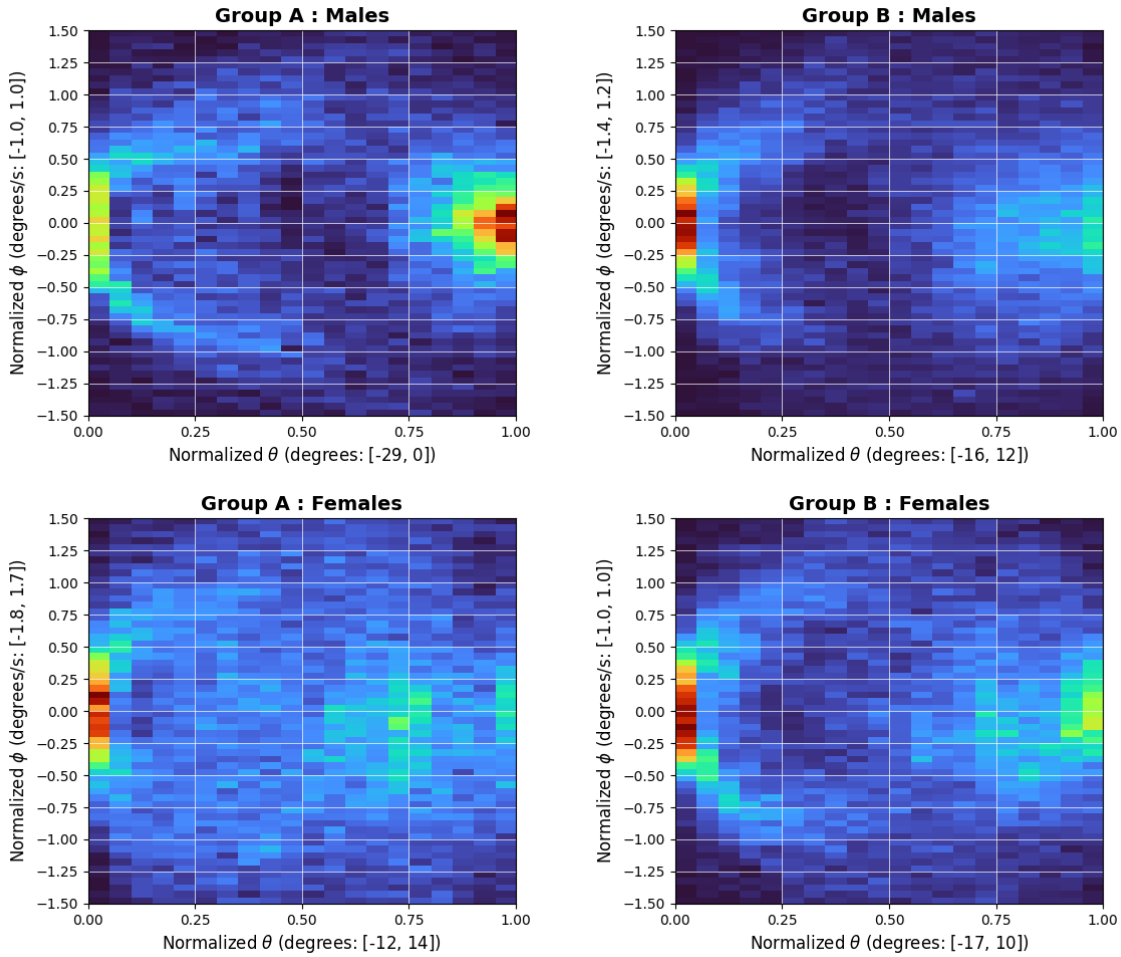


Figure 6: Characteristic state spaces of normalized lumbar angles (θ) and angular velocities (ϕ) for the bodyweight lunge, split by group and sex. Color scale indicates low frequency of state occurrence (blue) to high frequency of state occurrence (red).

No significant interaction or main effect of sex or group was found in the LNG UL condition for S, MC, and P-S test scores. However, a main effect of sex was found for E test scores ($F(1, 28) = 4.26, p = 0.04$), though no interaction or main effect of group was found. Pairwise comparisons showed that males had higher endurance times than females ($p = 0.04, r = 0.43$). For P-A test scores, there was no significant main effect of sex or group, however there was a significant interaction between sex and group ($F(1, 28) = 4.73, p = 0.04$). Pairwise comparisons showed that females had higher scores (less accuracy) than males in group B ($p = 0.02, r = 0.69$), but there was no difference between sexes in group A. Additionally, females in group B had higher scores (less sensitivity) ($p = 0.03, r = 0.67$) compared to females in group A. Results from the

ANOVAs are presented in Table 8.

Table 7: Lumbar capacity profiles for each movement group A and B of the bodyweight lunge, split by sex.

Group	Sex	N	P-A	P-S (^{o})	MC	E (s)	S (N)
A	M	9	0.23	10.84	-3.26	156.68	763.84
	F	6	0.20	12.02	-2.83	119.35	587.54
B	M	7	0.18	8.85	-2.78	150.56	775.26
	F	10	0.38	7.67	-2.86	103.04	613.56

Table 8: ANOVA test results comparing lumbar physical capacities across movement groups in the bodyweight lunge. Significant results ($p < 0.05$) are presented in bold with an asterisk.

Effect	P-A		P-S		MC		E		S	
	F	p	F	p	F	p	F	p	F	p
Sex	1.49	0.23	1.19	0.28	1.41	0.25	0.29	0.6	0.04	0.84
Group	<0.01	0.98	2.71	0.11	0.65	0.43	4.26	**0.04***	3.67	0.07
Sex × Group	0.21	0.65	4.73	**0.04***	1.71	0.20	0.06	0.81	0.01	0.94

Research question 2

Squat

The spectral clustering analysis produced two clusters of participants in the loaded SQT movement with a silhouette coefficient of 0.26, indicating a weak cluster strength. Group A (n = 19: 9M, 10F) had an average age of 21 ± 2 years, and an average BMI of 23.4 ± 3.2. Group B (n = 13: 7M, 6F) had an average age of 23 ± 3 years, and an average BMI of 23.2 ± 3.4. Average rate of perceived exertion given for loaded SQT movements was 4.3 ± 1.4, which corresponded to the descriptor of "Somewhat Hard". No differences in age, height, mass, or RPE were found between groups. Average state spaces and extracted lumbar capacity metrics for each group and sex are presented in Figure 7 and Table 9, respectively.

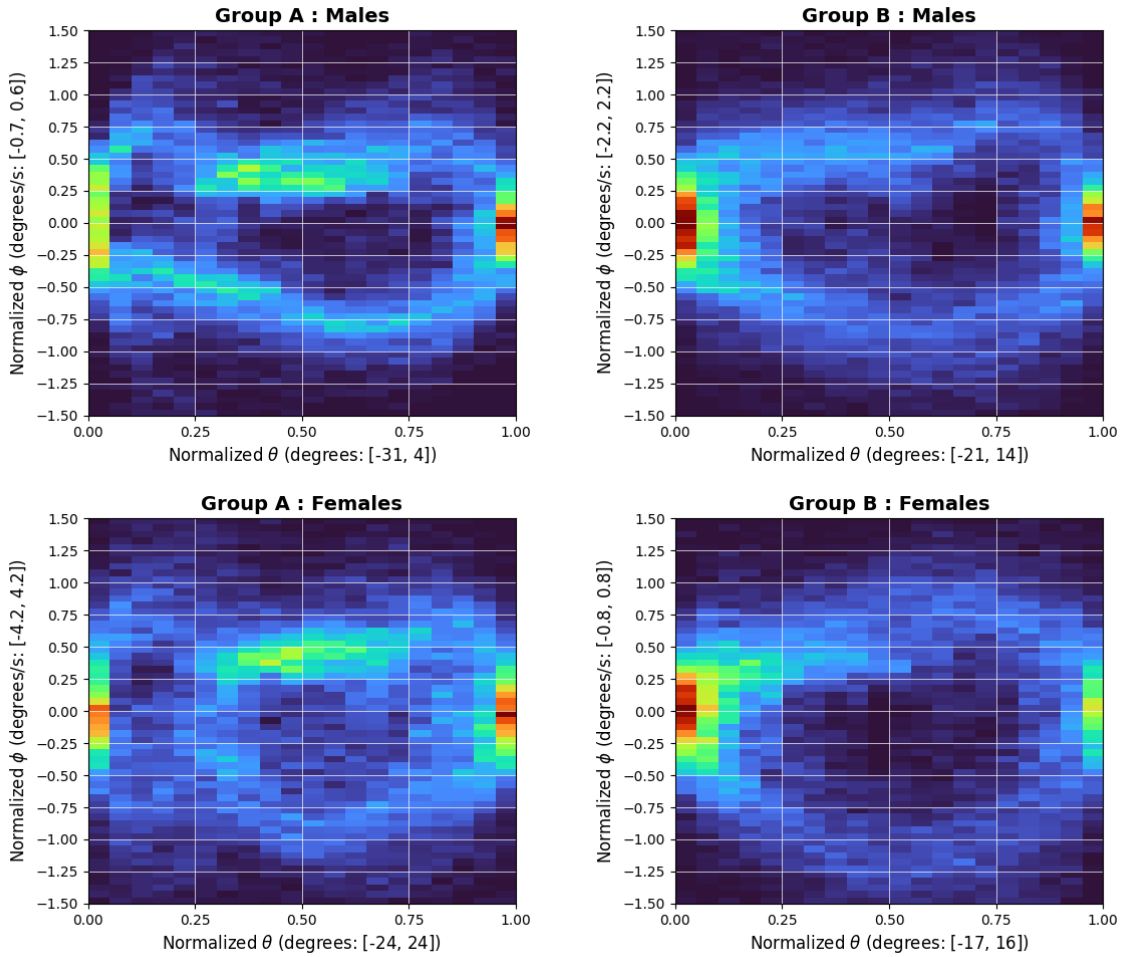


Figure 7: Characteristic state spaces of normalized lumbar angles (θ) and angular velocities (ϕ) for the loaded squat, split by group and sex. Color scale indicates low frequency of state occurrence (blue) to high frequency of state occurrence (red).

No significant interaction or main effect of sex or group was found in the SQT LD condition for S, MC, and P-A test scores. However, a main effect of sex was found for E test scores ($F(1, 28) = 5.31, p = 0.04$), though no interaction or main effect of group was found. Pairwise comparisons showed that males had higher endurance times than females ($p = 0.04, r = 0.43$). For P-S test scores, there was no significant main effect of sex or group, however there was a significant interaction between sex and group ($F(1, 28) = 7.91, p = 0.01$). Pairwise comparisons showed that males had higher scores (less accuracy) than males in group A ($p = 0.02, r = 0.64$), but there was no difference between sexes in group B. Results from the ANOVAs are presented in Table 10.

Table 9: Lumbar capacity profiles for each movement group A and B of the loaded squat, split by sex.

Group	Sex	N	P-A	P-S (^{o})	MC	E (s)	S (N)
A	M	9	0.23	12.05	-2.84	166.23	656.30
	F	10	0.37	6.17	-2.86	91.44	616.54
B	M	7	0.18	7.30	-3.32	138.28	913.53
	F	6	0.22	14.51	-2.82	138.69	582.57

Table 10: ANOVA test results comparing lumbar physical capacities across movement groups in the loaded squat. Significant results ($p < 0.05$) are presented in bold with an asterisk.

Effect	P-A		P-S		MC		E		S	
	F	p	F	p	F	p	F	p	F	p
Sex	0.49	0.49	2.87	0.10	1.34	0.26	0.20	0.66	2.02	0.17
Group	0.06	0.81	2.92	0.10	0.95	0.34	5.31	**0.03***	3.94	0.06
Sex × Group	7.91	**0.01***	0.83	0.37	1.75	0.20	3.69	0.07	3.23	0.08

Lunge

The spectral clustering analysis produced two clusters of participants in the loaded LNG movement with a silhouette coefficient of 0.33, indicating a weak cluster strength. Group A (n = 13: 10M, 3F) had an average age of 22 ± 3 years, and an average BMI of 23.6 ± 3.4. Group B (n = 19: 6M, 13F) had an average age of 21 ± 2 years, and an average BMI of 23.1 ± 3.2. Average rate of perceived exertion given for loaded LNG movements was 3.7 ± 1.3, which corresponded to the descriptor of "Somewhat Hard". No differences in age, height, mass, or RPE were found between groups. Average state spaces and extracted lumbar capacity metrics for each group and sex are presented in Figure 8 and Table 11, respectively.

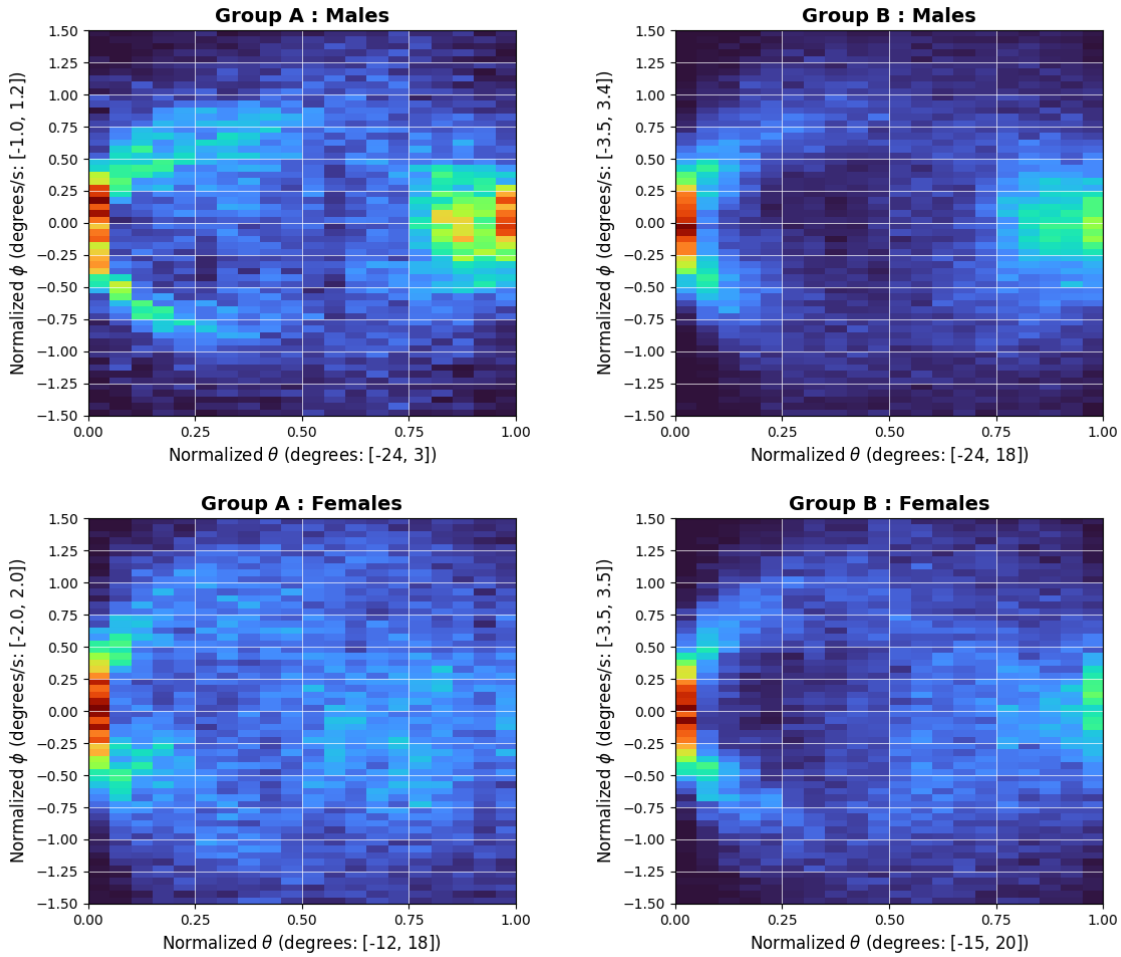


Figure 8: Characteristic state spaces of normalized lumbar angles (θ) and angular velocities (ϕ) for the loaded lunge, split by group and sex. Color scale indicates low frequency of state occurrence (blue) to high frequency of state occurrence (red).

No significant interaction or main effect of sex or group was found in the LNG LD condition for S, E, and P-A test scores. For MC test scores, there was no significant main effect of gender or group, however there was an interaction between gender and group ($F(1, 28) = 4.93, p = 0.03$). Pairwise comparisons showed that females had higher scores than males in group A ($p = 0.03, r = 0.87$), but there was no difference between genders in group B. Additionally, group B males had higher scores than group A males ($p = 0.04, r = 0.63$). For P-S test scores, there was no significant main effect of gender, however there was a main effect of group ($F(1, 28) = 4.67, p = 0.04$), and a significant interaction between gender and group ($F(1, 28) = 4.24, p = 0.04$). Pairwise comparisons for the main effect of group showed no significance. However, pairwise comparisons

for the interaction showed that females in group A had higher scores (less sensitivity) ($p = 0.04, r = 0.79$) compared to females in group B. Results from the ANOVAs are presented in Table 12.

Table 11: Lumbar capacity profiles for each movement group A and B of the loaded lunge, split by sex.

Group	Sex	N	P-A	P-S (^{o})	MC	E (s)	S (N)
A	M	10	0.22	10.46	-3.25	158.62	781.15
	F	3	0.22	19.17	-2.52	111.95	574.46
B	M	6	0.19	9.16	-2.70	146.31	748.32
	F	13	0.33	7.02	-2.92	108.51	610.58

Table 12: ANOVA test results comparing lumbar physical capacities across movement groups in the loaded lunge. Significant results ($p < 0.05$) are presented in bold with an asterisk.

Effect	P-A		P-S		MC		E		S	
	F	p	F	p	F	p	F	p	F	p
Sex	4.67	0.04	0.22	0.64	0.68	0.42	0.14	0.71	<0.01	0.95
Group	0.49	0.49	1.84	0.19	0.37	0.55	3.23	0.08	2.84	0.10
Sex × Group	4.24	**0.04***	0.99	0.33	4.93	**0.03***	0.03	0.85	0.12	0.73

A summary of the significant findings is in Table 13 below.

Table 13: Summary of significant results.

Movement	Condition	Capacity	Result	p-value	Effect size
Squat	Bodyweight	P-S	Group A: Males > Females	0.04	d=1.17
			Males: Group A > Group B	0.02	d=1.04
		E	Males > Females	0.03	r=0.43
	Loaded	P-S	Group A: Males > Females	0.02	r=0.64
		E	Males > Females	0.04	r=0.43
Lunge	Bodyweight	P-A	Group B: Females > Males	0.02	r=0.69
			Females: Group B > Group A	0.03	r=0.67
		E	Males > Females	0.03	r=0.43
	Loaded	P-S	Females: Group A > Group B	0.04	r=0.79
		MC	Group A: Females > Males	0.03	r=0.87
			Males: Group B > Group A	0.04	r=0.63

Meta-clustering of movement profiles

The spectral clustering analysis on common movement groups produced five clusters of participants with a silhouette coefficient of 0.75, indicating a strong cluster strength (Figure 9). ANOVA results indicated no significant differences between the meta-clusters in terms of age, height, mass, or lumbar capacity scores (Table 14). One-sample Chi-squared tests showed that two of the five groups were predominantly male or female (Table 15).

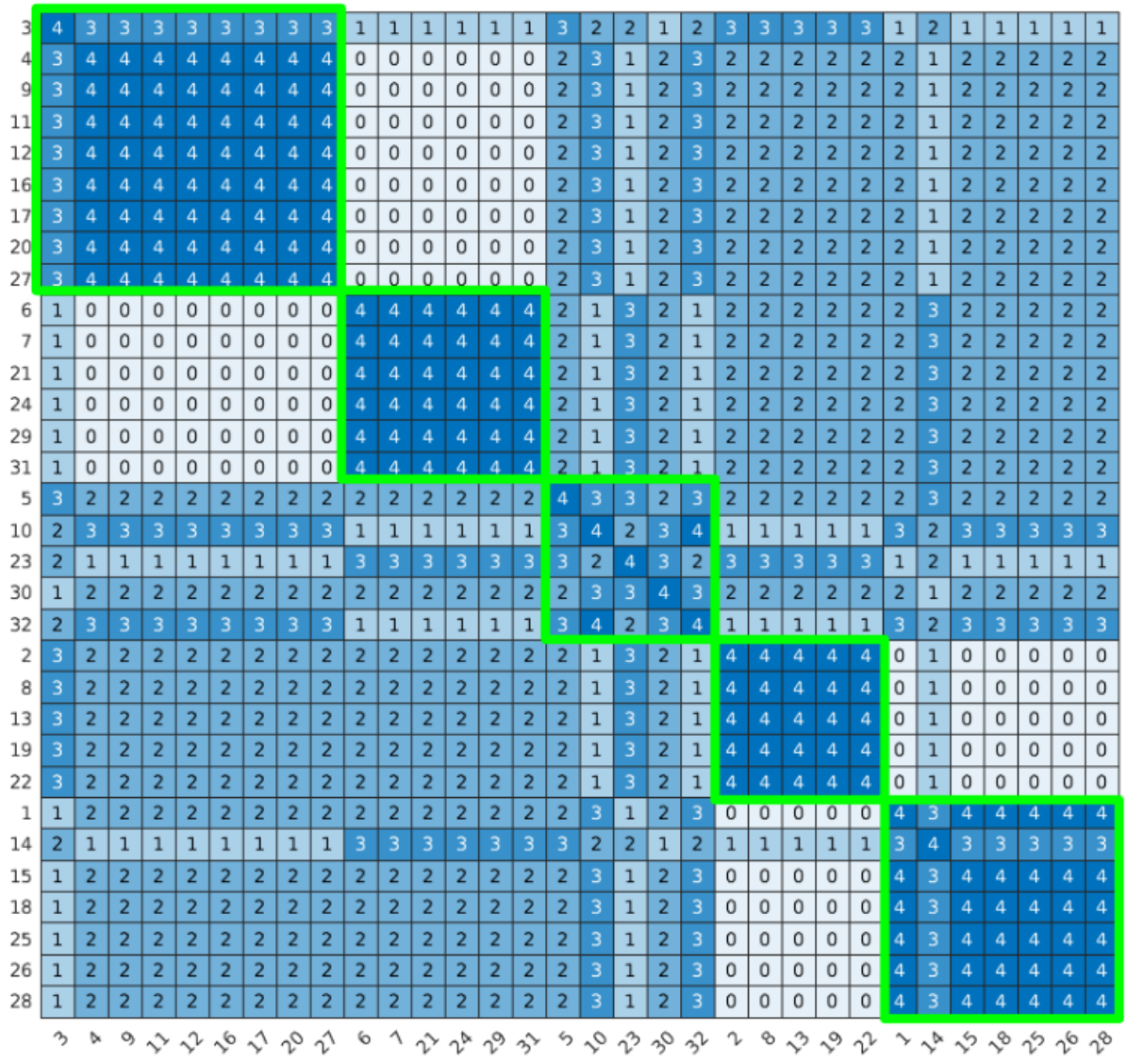


Figure 9: 32 X 32 matrix based on the similarity of bodyweight/loaded squat/lunge movement groups for each participant. Cells are shaded based on the number of common movement groups (0-4). Green blocks highlight the 5 detected meta-clusters from spectral clustering.

Table 14: ANOVA test results comparing lumbar physical capacities across meta-movement clusters.

	Age	**Height**	**Mass**	**P-A**	**P-S**	**MC**	**E**	**S**
F	0.97	2.73	2.65	1.57	2.43	0.77	0.38	0.76
p-value	0.44	0.05	0.06	0.21	0.07	0.56	0.82	0.56

Table 15: Summary of meta-movement clusters and their composition. Significant sex differences are highlighted in bold with an asterisk

Meta-cluster	**Sex**		χ^2	**p-value**	**Movement Groups**			
					Squat		**Lunge**	
	M	**F**			**Bodyweight**	**Loaded**	**Bodyweight**	**Loaded**
I	4	1	6.92	**<0.01***	A	A	A	A
II	3	4	0.53	0.47	B	B	B	B
III	2	7	10.8	**<0.01***	A	A	B	B
IV	4	2	2.47	0.11	B	B	A	A
V	3	2	0.91	0.34	B	A	Mixed	Mixed

Chapter V: Discussion

Rank correlation coefficients showed overall no correlations between capacities with the exception of two medium-low negative correlations between P-A and P-S and S and E. This indicates is that there was no distinct high-low ranking of participants according to capacity test scores, and justifies the more nuanced clustering based approach to form movement groups.

The results of clustering analyses on lumbar dynamics in the SQT and LNG confirmed the presence of heterogeneity in the sample population. Two subgroups were found in each of the movement tasks and loading conditions. While silhouette coefficients only indicated weak strength of clustering, visualizations of the state spaces highlight clear pattern differences between groups. State spaces in the SQT movement mainly highlighted differences in shape (i.e., narrowing and expanding in different areas, and shifting of attractors) while state spaces of the LNG mainly highlighted differences in diffuseness of the space, and intensity of attractors. Likewise, loading conditions mainly produced differences in location and intensity of attractors. In relating state spaces to movement patterns, diffuseness versus high intensity attractors can be thought of as representing timing and variability of movement patterns, while shape changes can be thought of as movement pattern differences. Additionally, it is important to note that higher velocity of movement causes fewer data points in a particular state for the trajectory, since the system is moving more quickly in and out of that state. Differences in some of the lumbar capacities were found between clustered movement groups and as such, the specific features of movement state spaces distinguishing these capacities are addressed in the following discussion.

In both SQT and LNG state spaces, two clear attractors are present: one at the highest lumbar extension angle, and another at the highest lumbar flexion angle, both of which correspond to the top and bottom movement positions, where the body is transitioning between ascent and descent. In the LNG, these are the only attractors, while in the SQT, there are visible attractors between these two states. This is understandable based on the nature of the two movements, with the SQT typically utilizing

a larger range of lumbar motion, and higher peak flexion angles compared to the LNG, consistent with previous literature [Kingma et al., 2010, Bayartai et al., 2020]. Thus, larger ranges of motion in the SQT produced the possibility of a more complex shape, while the smaller range of motion in the LNG produced higher intensities in the state spaces but less complex shapes. In general, the state space shapes of LD conditions are fairly similar to the UL conditions for the respective movement. However, some differences are interesting to note. In particular, UL conditions demonstrated slower dynamics in the upper range of the movement, while LD conditions demonstrated a shift to slower dynamics in the lower range of movement, as represented by the higher intensities in the right versus left regions of the state space, respectively. The slower dynamics in the lower range of movement with load could be related to prior findings that when comparing relative phase between the knee joint and lumbar spine in a lifting task, higher loading increased the lag of lumbar spine extension behind the knee in the lifting phase [Scholz, 1993].

In three of the four grouped conditions (SQT UL, SQT LD, and LNG UL), males had a higher endurance score than females. Interestingly, for the SQT conditions, characteristic phase spaces showed a more even intensity in the lumbar flexion (descending) and extension (ascending) phases of the squat compared to females, who spent more time at the top, bottom, and lowering phases of the squat. Since eccentric muscular contractions are reported to have greater contribution to force through passive mechanisms [Herzog, 2014], it is likely that in a fatigued state, an individual will take more time in the eccentric phase than the concentric phase. While this study did not compare between unfatigued and fatigued states, it can be assumed that some fatigue did show up in the latter repetitions of each set, which would influence the trajectory of lumbar dynamics sooner in individuals who exhibited less endurance capacity.

In both conditions of the LNG movement, it is clear that two main attractors are present: one in the maximal extension, and another in the maximal flexion of the lumbar spine during the movement. Since the intensity of the trajectories between these two points is very low and diffuse, this indicates that the LNG produced a quick shift

from a more extended posture to a slightly flexed posture, likely associated with the front leg load bearing phase of the movement. In a more fatigued state, it could be expected that an individual will transition, or "collapse", more quickly to the flexed posture, and be unable to maintain as much of a lordotic curve [Sparto et al., 1997, Weeks et al., 2015]. This can explain why the attractor is stronger in the initial phase for males compared to females in the LNG UL condition. Higher endurance would allow the males to maintain the more extended lumbar posture for longer. While this effect is also visible in the trajectories of the LNG LOAD condition, the contrast between males and females is less, which may explain the lack of significant difference of endurance between sexes.

An interaction between group and sex for P-S was found in three of four conditions (SQT UL, SQT LD, and LNG LD), however, the implication and relevance of these results to the trajectory of the lumbar state spaces is not clear.

In the SQT UL, group B males had better sensitivity (lower test score indicating less error) compared to group A males. A very clear difference in averaged trajectory of lumbar dynamics is in the narrowing (slowing velocity) versus expanding (increasing velocity) of the lumbar angle between the two groups. The increased range in the lumbar angular velocity values for Group A compared to group B indicate the higher level of fluctuation in the movement pattern of Group A males.

Group A can be described as having an initial wobble slowly moving towards maximal flexion, which then quickly collapses in the lower end of the movement. Conversely, group B can be described as having a faster initial movement into flexion, which becomes slow and controlled at the bottom range of motion. Slowing down of flexion towards the bottom of the squat could be because of a controlled slowing in response to the sensation of a collapse into flexion (and thus a compensatory response to try to keep a lordotic curve, as is commonly instructed and trained for in squat movements). This may explain the higher sensitivity capacity seen in group B males compared to group A males.

Interestingly, however, group A females also had better sensitivity than group A

males. In this case, the mid-lowering and bottom attractors for females had a more intense profile compared to males, which may also relate to a reduced amount of maximal flexion experienced by the females. The intensity of the attractors indicates a slowing of movement in the corresponding phases of motion, which similar to above, could be in response to a sensation of lumbar flexion. However, this pattern of narrowing of the trajectory is seen in group B females compared to group A females while no significant difference in sensitivity was found between the two groups, so it is unlikely that this narrowing can be explained fully by a compensatory response due to higher position sensitivity capacity. The SQT LD condition also showed significantly better position sensitivity in group A females compared to males, as well as in group A females compared to group B females. As previously noted, a slight narrowing in the group A female that is not evident in the other groups can perhaps explain some of the higher sensitivity, however, the pattern is not very clear and likely does not fully explain the group sensitivity differences.

The LNG LD condition showed that females in group B had better sensitivity than females in group A. Because the LNG movement is very different from the SQT movement, narrowing as discussed earlier for the SQT is not expected, and is indeed not present. However, a clear difference in diffuseness of the averaged trajectories is very clear when comparing these two groups, with the higher sensitivity group (B) showing a much more consistent trajectory, compared to the large variability of the lower sensitivity group (A). It is possible that low position sensitivity was the cause of more variant patterns, even within-participant, compared to individuals with better position sensitivity.

While the implications of both P-S and P-A scores on movement patterns are not very clear, this study showed that P-S and P-A are not the same in that different effects were found in groups based on different movement patterns. This agrees with the conclusions made in prior research [Lowrey et al., 2020] that movement discrimination and joint matching tasks are not the same thing. Rather, they should be treated as different aspects of proprioception.

Diffuseness in the state space could also be related to aspects of motor control. In the LNG LD condition, motor control was higher in group A females compared to males. The more diffuse, lighter intensity seen in the group A female state space could be explained by the fact that smoother movement would mean that there is equal time spent in each "bin" of the state space. Rather than only a quick shift between the two postures, there is more of a smooth transition between postures. The higher MC seen in group B males compared to group A males can also be explained the same way (more diffuse intensities in group B males). Differences in MC could also explain the higher intensity in the attractor at the top of the movement seen in the state space for group A males. In essence, this represents more time spent in the attractor at the top of the LNG movement, and less smoothness in transition between the two attractor states. Interestingly, a similar attractor at the top of the movement is seen in the LNG UL condition for the group A males, where no effect of group or sex was found for MC. This attractor is more constrained than in the LNG LD condition (where the interaction of group and sex on MC was found), so it is possible that this explains some of the effect difference.

In all movement and loading conditions, no effect or interaction of sex and group were found for S. This is despite the fact that sexes had significantly different S scores, as confirmed by a t-test. However, the movement profiles produced a fairly even split between males and females for all conditions (aside from LNG LD, in which all but three females were in group B). It is interesting that S differences did not have much bearing on movement patterns, though it is likely that the moderate loading in the LD conditions was not enough to produce a strong enough challenge to individuals, whereby differences in strength would be highlighted.

The analysis techniques used in this study (i.e., state space construction and spectral clustering) were different from techniques used previously in biomechanics literature. Characterization of movement patterns has typically been done using vector coding or continuous relative phase (CRP) analyses [Zehr et al., 2018, Maddox et al., 2020, Needham et al., 2014]. Both of these techniques compare between two joints, as

opposed to a single joint's state (i.e., angular position and velocity) and trajectory as was done in this study. Vector coding uses a state space technique (plotting the position of two joints against each other), however the interpretation of vector coding simplifies the data such that movement is represented as being binned into categories based on the angle between states in consecutive time points. As such, information about the movement pattern is lost, in particular for more nuanced differences between patterns. For example, vector coding was able to distinguish movements of different lifts (flexed versus neutral spine lifting techniques), but could not distinguish between either of these or a freestyle lift, in which overall technique differences were less obvious [Zehr et al., 2018]. The technique in the current study allows for a comparison of movement patterns within the same task and provides insight into more nuanced differences. Clustering of individuals based on movement patterns has also previously utilized a K-means clustering technique [Beaudette et al., 2019]. This technique clusters data points based on global connectivity (i.e., distance between global data positions) and thus groups data points within a certain radius of a centroid. In contrast, spectral clustering distinguishes the local connectivity of nearest neighboring data points (defined by eigenvalues of the similarity matrix) from global connectivity and thus is more robust in data sets where clusters may produce non-spherical shapes.

An exploratory investigation into how individuals grouped across all combinations of movement and loading conditions prompted the meta-clustering analysis. Interestingly, this meta-clustering was able to produce five clusters, with strong cluster strength. This may be an indication that clustering individuals on combinations of movement classifications rather than movement patterns in single tasks has the potential to classify individuals better. Intuitively, this makes sense since the combination of performances in multiple tests is more comprehensive in characterizing an individual's behavior. For example, meta-clustering may be able to distinguish features within an individual's movement that are present consistently (e.g., indicating a trait), from features that show up only sometimes or in certain situations. However, meta-clustering may also make it more difficult to interpret results and differences (or lack of differ-

ences) between groups. For example, no significant differences in age, height, mass, or lumbar capacity scores were found between these groups, though it was interesting to note that two of the five groups had a significantly different proportion of male and female participants.

A lack of differences in lumbar capacities found for the meta-groups does not exclude the potential for implications to the relevance of capacities on combinations of movement classifications. Rather it simply indicates that the SQT and LNG movements in UL and LD conditions may not be different enough in their challenge to each capacity (e.g., they may not be distinct enough from each other in the way they target or challenge each of the capacities). While it is difficult to make conclusions based on these meta-clustering results, a more comprehensive analysis with more movement tasks (in particular, tasks that are more distinct in goal and challenge) as well as a higher sample size may be warranted.

Chapter VI: Conclusion

Research implications

The main implications for this research are in uncovering fundamental aspects of lumbopelvic movement during the SQT and LNG that differ in subgroups of individuals with distinct movement strategies and relating them to physical low back capacities of strength, endurance, proprioception, and motor control. Using state spaces to characterize the dynamics of the lumbar spine has proven helpful in allowing for a more complex and nuanced understanding of the movement.

Characteristic state spaces of lumbar dynamics between groups can be distinguished by their overall shape and diffuseness as well as intensity and location of attractors. Scores of E and P-S were typically different between clusters, highlighting that endurance and joint position sensitivity in particular may be more influential on lumbar spine dynamics in whole-body functional movements, both in unloaded and moderately loaded conditions. As such, this study suggests that improving these capacities through training may be a means of improving movement patterns and reducing the risk of injury (e.g., promoting control of the lumbar spine, particularly in flexed postures).

The results of this study also suggest that it may be possible to infer an individual's current strengths and weaknesses based on their movement patterns, which has implications for individualization of training, progress assessment, and rehabilitation from injury by informing specific, targeted interventions. Further research is required to parse out the exact relationships between the specific capacities and their effects on movement, as well as to investigate whether individuals transition between movement subgroups with targeted training. Furthermore, this work has implications for future investigation into the assessment of compensatory movement and early recognition of the pathological development of movement disorders.

Limitations & future directions

One main limitation of this study is the sample size. In particular, a larger sample size could have strengthened the cluster silhouette scores, and the ability to detect subgroups at a more granular level. It is also likely that the sample population did not fully encompass the range of movement profile subgroups that would cluster in the larger population. However, the weak clustering produced from these data allowed for meaningful comparisons to be made.

Other factors that may have limited the results of this study include spatio-temporal movement constraints necessary for the standardization of the functional tasks may have influenced the natural movement patterns of participants, potentially masking or creating differences in subgroups of the sample. For example, controlling for the velocity of movement may have masked some of the heterogeneity within the population that would come about through natural variation in movement speeds. However, for the purpose of this study, it was necessary to control for the speed of movement to allow for better comparisons within the movement tasks, since it is known that speed influences the movement strategy [Frost et al., 2015b]. Future work should investigate the heterogeneity that is apparent with the additional factor of movement speed in each task. Another limitation was that the use of bodyweight to standardize loaded conditions did not account for body composition and overall relative strength of participants, which may have led to different levels of intensity experienced. While self-perceived intensity based on RPE was consistent across participants and confirmed moderate intensity of loaded conditions, a more rigorous investigation into the effect of load on movement patterns is warranted, with better matching of load to participant ability, as well as greater variation of load intensities.

Further understanding of how individuals behave between repetitions of a movement task is also required. The current study collapsed an individual's state spaces in each condition so that an overall characteristic trajectory could be used for clustering. Future work could cluster on individual repetitions of movement and characterize individuals based on a percentage of repetitions completed in different groups (if it is the

case that they are not perfectly consistent in their behavior). This would also bring insight into which features of movement can be considered a trait of the individual, rather than a combination of other internal and external factors and circumstance. It would additionally allow for an investigation of fatigue, by looking at later repetitions compared to early repetitions of the movement.

Additionally, lumbar capacities were characterized by tests that do not fully represent the associated capacity. For example, motor control was only assessed using a motion smoothness metric, while other aspects, such as reflexive control were not represented in the motor control capacity. Motion smoothness was selected as the representative motor control test for the purpose of this study since it has been used previously [Perrott et al., 2021, Miller et al., 2023], could be isolated to the low back, and did not require extensive equipment (such as a robotic platform) as used in other tests [Reeves et al., 2014, Reeves et al., 2018]. Likewise, proprioception only assessed accuracy and sensitivity in a very narrow range of lumbar angles, which may not reflect an individual's proprioceptive ability across the entire range of motion. Furthermore, the cat-camel approach (with participants on all fours for both P-A and P-S tests) used in this study has not been validated in the literature, although the concept was taken from previous work. Specifically, the joint repositioning accuracy (letting participants gauge their full range then find the halfway position) as well as the joint position sensitivity (measuring the threshold at which a change is detected) have both been used previously [Swinkels and Dolan, 1998, Lee et al., 2010], though in different positions. The cat-camel test allowed for these proprioceptive capacities to be tested without extensive equipment such as a motorized apparatus with a moving platform base. However, further work validating the cat-camel as a method of testing these distinct proprioceptive capacities is required. A larger battery of tests, or a more comprehensive metric for each of the individual capacities would be beneficial to confirm and strengthen the observations made. Including more specific capacity tests in the future could also lead to better translation into clinical and occupational practices.

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