

Effects of posture and trunk muscle coordination on multi-joint isometric lifting strength: Implications for individualised movement assessment and intervention

by

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This thesis consists of material all of which I authored or co-authored: see Statement of Contributions included in the thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Statement of Contributions

Brendan Pinto was responsible for conception and led study design, data collection, analysis and writing of this thesis and the manuscripts produced from this thesis.

Dr. Jack Callaghan contributed to conception, study design, interpretation, writing of the manuscripts produced from this thesis and provided feedback on this thesis. He also secured external funding for the research and provided laboratory equipment.

Dr. Tyson Beach contributed to conception, study design, interpretation, writing of the manuscripts produced from this thesis and provided feedback on this thesis.

Dr. Samuel Howarth contributed to data analysis, interpretation, writing of the manuscripts produced from chapter 5 of this thesis.

Abstract

The manner in which movement is executed can influence biomechanical demand and consequently the development of musculoskeletal disorders. Although modifying movement execution can help regulate biomechanical demand by influencing tissue loading and tolerance, it can also influence physical performance by influencing the ability to exert force. Movement-based interventions to regulate biomechanical demand that hinder physical performance can limit acceptability, sustainability, and effectiveness of the intervention. Conversely, interventions to enhance physical performance that modify movement execution in a way that imposes higher than necessary biomechanical demand, can hinder physical development and long-term performance. However, the development of comprehensive movement assessments and interventions that consider both the impact on biomechanical demand as well as physical performance is challenged by a scarcity of knowledge of how movement execution can influence physical performance in multi-joint tasks. The global aim of this thesis was to investigate how modifying posture and trunk muscle coordination can influence the ability to exert force during isometric lifting.

Four studies were conducted to address the global thesis aim. Study 1 investigated the effects of modifying trunk inclination and low back curvature on isometric lifting strength as previous research has yet to clearly dissociate how these distinct postural characteristics could influence the ability to exert force in multi-joint tasks such as lifting from the ground. Results showed that modifying trunk inclination and low back curvature can influence strength in multi-joint tasks substantially and to a similar extent. However,

the effect of modifying these postural characteristics interact and vary greatly across individuals in magnitude (up to 620N of isometric lifting force) and direction (increase/decrease). Thus, a single postural profile cannot be generalized as the strongest for every individual and the ability to exert force in multi-joint tasks cannot be inferred solely from posture. Additionally, the individually varying effects of posture on strength suggests that individuals can adapt through movement training to be stronger in postures that are favourable for tissue loading and tolerance. Assessing the L4/L5 joint contact forces of the observed postures suggested that the effect of posture on biomechanical loading is more consistent, and the biomechanical demand imposed by flexing the low back outweighs any potential acute gain in isometric lifting strength. Together this supports the recommendation to avoid highly flexed low back postures during demanding physical activities.

Study 2 compared the immediate effects of a simple verbal directive and detailed trunk muscle bracing coaching on isometric lifting strength, low back postural displacement and trunk muscle co-contraction. Prior research has suggested that cueing co-contraction of all the trunk muscles can enhance the ability to exert force in multi-joint tasks but has not yet isolated the effect of modifying trunk muscle coordination on the ability to exert force which may potentially depend on the approach used to cue trunk muscle coordination. Detailed coaching which included a combination of verbal and physical cues was more effective than the simple verbal directive at increasing trunk muscle co-contraction (group mean co-contraction for the baseline, directive and coached condition was 10.1%, 11.0% and 13.5% respectively) and decreasing low back postural displacement (group mean change in low back flexion angle normalized to each

individual's maximum flexion range-of-motion for the baseline, directive and coached condition was 21.4%, 19.5% and 17.2% respectively). However, both cueing approaches immediately reduced isometric lifting strength to a similar extent (group mean peak isometric lifting force for the baseline, directive and coached condition was 1194 N, 1109 N and 1096 N respectively). Results indicate that detailed coaching is more effective than simple verbal directives at modifying trunk muscle coordination to restrict low back postural displacement but cautions cueing trunk muscle coordination for the first time in situations where peak force production is desired. Results also suggest that future research should confirm acquisition and transfer of the targeted trunk muscle coordination patterns as the full potential impact of modifying trunk muscle coordination may not be completely apparent from observing the immediate responses to cues.

Motivated by the individually varying effects of posture on the ability to exert force observed in Study 1, Study 3 evaluated the extent to which proxies for leverage derived from kinematic quantities can statistically explain the individual variation in the effects of posture on the ability to exert isometric lifting force. Prior research has used kinematic-based proxies to describe and make inferences about how posture can influence leverage in multi-joint tasks. However, these approaches do not capture all the mechanics involved in multi-joint kinetics and the extent to which they may explain the individually varying effects of posture on the ability to exert force in a multi-joint task has not yet been tested. As expected, based on fundamental biomechanical principles, the kinematic-based proxies for leverage that were investigated (joint-to-external force moment arms and predicted joint-angle-dependent torque-generating capacities) explained a very low proportion of variation (<17%) in the effects of posture on isometric lifting force. In

contrast, variables derived from both kinematic and kinetic measurements such as the net joint reaction moments calculated using a rigid linked segment model and inverse dynamics explained a higher proportion of variation in the effects of posture on the ability to exert isometric lifting force (approximately 80%). These results indicate that simplified kinematic-based approaches cannot be used to assess the effects of posture on leverage in multi-joint tasks on an individual basis. Instead, variables derived from both kinematic and kinetic measurements such as the net joint reaction moments show promise for being used in development of quantitative assessments of multi-joint leverage.

Study 4 investigated whether individuals maintain their potential for physical performance when given instruction to avoid rounding the low back during light mass lifting. Compared to prior movement-based interventions such as the squat lift technique that vaguely describes whole body posture, targeting a key postural feature such as low back flexion is theoretically expected to afford greater flexibility to self-organize the rest of the body linkage to regulate biomechanical demand, without hindering physical performance. However, using simple verbal directives to cue specific movement features during low demand tasks may not acutely prompt individuals to prioritize physical performance as they self-organize, rendering the intervention ineffective. Although aggregate group level results indicated that low back postural instruction targeting biomechanical demand decreased low back flexion during crate lifting and increased isometric lifting strength in postures replicating those exhibited during crate lifting, there was high heterogeneity in responses. Among the 37 of 40 participants classified as individuals who could potentially benefit from decreasing their low back flexion, 15 and 22 participants were respectively classified as successful and unsuccessful in decreasing

low back flexion in response to the instruction, to be within a range that minimizes passive tissue strain. Though replicating the crate lifting posture to assess isometric strength emerged to be challenging, there were individuals who increased (n=8), decreased (n=9) and did not change (n=23) isometric strength, independent of their low back postural response. Hence, although most participants appeared to maintain physical performance potential when given simple verbal directives that target low back flexion to regulate biomechanical demand, some participants responded in a manner that decreased performance potential and many did not successfully decrease their low back flexion within a range that minimizes passive tissue strain. This suggests a need for more detailed movement coaching and training in research and practice to effectively modify movement behaviour without hindering physical performance potential.

Overall, the results from this thesis indicate that the effects of modifying posture and trunk muscle coordination on the ability to exert force in multi-joint tasks is complex and can greatly vary across individuals. Additionally, modifying movement execution may need to go beyond simple verbal directives to provide detailed movement coaching. The findings support that considering the effects of movement execution on physical performance has the potential to advance movement assessment and intervention strategies. Yet, there is a need to develop approaches to capture the individually varying effects of movement features such as posture on the ability to exert force in multi-joint tasks to develop strategies that can effectively regulate biomechanical demand without hindering physical performance as well as enhance physical performance without imposing unnecessary biomechanical demands.

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Chapter 1 : General Introduction

The way we execute movement has been linked with low back disorders (LBD; Marras, 2008; Marras et al., 2016; McGill, 2009, 2016), which continues to be the global leading cause of disability (Vos et al., 2016; Wu et al., 2020) and creates enormous health care and worker disability costs (Hartvigsen et al., 2018). Specifically, the motions, postures, and forces of the movement can influence biomechanical demand which encompasses tissue loading and tissue tolerance, to influence risk of injury and perception of pain (Cholewicki et al., 2019; Marras, 2008; Marras et al., 2016; McGill, 2009, 2016; Nelson & Hughes, 2009). For this reason, solely reducing the external task demands (e.g. mass lifted) does not necessarily reduce biomechanical demand and risk of developing LBD, as the way in which individuals move have the potential to counteract task modifications (Faber et al., 2007; McGill, 2009). This knowledge has motivated assessing and modifying movement features such as trunk posture (Beach et al., 2018; Chan et al., 2022; Frost et al., 2015; Gagnon et al., 2016; Pinto et al., 2018; van Dieën et al., 1999) and trunk muscle coordination (Huxel Bliven & Anderson, 2013; Liebenson et al., 2009) to manage biomechanical demand. However, solely focusing on the effects of biomechanical demand may be inappropriate as modifying trunk posture or muscle coordination may also influence physical performance.

Physical performance is not just relevant in contexts where the objective is peak performance such as athletic endeavours. For example, a squat lift technique to maintain a vertical trunk inclination has been promoted for decades in manual materials handling training research and practice (Denis et al., 2020; van Dieën et al., 1999) and continues

to be perceived as the safest technique across untrained individuals, trained individuals, and manual materials handling trainers (Abdoli-Eramaki et al., 2019). However, adopting a squat technique when lifting from the ground can decrease isometric and isokinetic lifting strength (Kumar & Garand, 1992), increase energy expenditure (Garg & Herrin, 1979; Kumar, 1984; Welbergen et al., 1991), be perceived as more exerting (Straker & Duncan, 2000) and reduce the ability to balance (Chow et al., 2005; Heiss et al., 2002). These factors can hinder task performance and have been suggested to be responsible for the poor adherence and adoption of the squat lift technique by individuals (Denis et al., 2020). Thus, understanding how movement execution can impact physical performance is also of interest in submaximal tasks where the objective is not peak physical performance.

Previous research has focused on how modifying posture or trunk muscle coordination can influence biomechanical demand, but less attention has been given to the effects on physical performance. For instance, the effects of posture on the ability to exert force at the end-effector during multi-joint tasks has primarily been investigated by comparing postural techniques that encompass vaguely defined whole-body postural profiles. The squat and stoop lift (Kumar & Garand, 1992), and the conventional and sumo deadlift (Cholewa et al., 2019; Cholewicki et al., 1991; Escamilla et al., 2000) are examples of techniques that have been investigated to determine the effects of posture on lifting strength. However, investigating postural techniques has limited inferences that can be made about how to modify movement execution as the execution of these techniques can vary based on each individual's interpretation of the given instruction (Abdoli-Eramaki et al., 2019) and capacity to perform the movement (Beach, Frost, Clark,

et al., 2014; Li & Zhang, 2009; Patterson et al., 2022), which results in individual differences in posture. This variation contributes to the conflicting results reported by prior studies which has made the impact of modifying posture on the ability to exert force unclear as some show an effect (Kumar & Garand, 1992) whereas others do not (Cholewa et al., 2019). Technique comparison studies have not typically statistically accounted for or experimentally controlled for individual differences in specific postural features such as trunk inclination and low back curvature, challenging a conceptual framework for how posture can influence the ability to exert force.

Inferences made from prior research have also led to the emergence an interesting paradox which suggests that peak physical performance requires postures that impose higher biomechanical demands. Specifically, while low back flexion can increase tissue loading (Potvin et al., 1991; Zwambag & Brown, 2020) and reduce the tolerance of functional spinal units to loading (Callaghan & McGill, 2001; Gallagher et al., 2005; Gunning et al., 2001; Howarth & Callaghan, 2012), back extensor strength appears to increase with trunk flexion which is a combination of trunk inclination (i.e. hip flexion) and low back flexion (Daggfeldt & Thorstensson, 2003; Gallagher, 1997; Graves et al., 1990; Kocjan & Sarabon, 2014; Tan et al., 1993). This highlights a further challenge in the literature, which is that there has often been a lack of a clear distinction between trunk inclination and low back curvature. Although it remains unclear if increases in back extensor strength with trunk flexion are due to trunk inclination or low back curvature, back extensor strength alone does not necessarily indicate how force exerted at the end-effector during multi-joint tasks such as the hands during lifting, will be impacted by trunk posture. It is necessary to investigate if this paradox between biomechanical demand and

physical performance indeed exists in the physical performance of multi-joint tasks, to appropriately guide strategies to enhance physical performance and development. For instance, if efforts to increase physical capacity such as exercise training are performed with movement execution that imposes higher than necessary biomechanical demand to increase acute physical performance based on the above-mentioned paradox, the resulting adaptations to training can be disadvantageous by increasing the risk of developing pain or injury.

Apart from the effects of posture on the ability to exert force in multi-joint tasks, trunk muscle coordination has been suggested to improve force exertion at the end-effector in multi-joint tasks (Kibler et al., 2006; McGill, 2010, 2011). Training the trunk muscles has been advocated as an important component to low back health as well as physical performance during daily living and athletic endeavours (Granacher et al., 2013; Huxel Bliven & Anderson, 2013; Saeterbakken et al., 2022). However, the effects of cueing trunk muscle coordination to influence the ability to exert force at the end-effector has only been speculated from observing the behaviour of elite strength athletes (McGill, McDermott, et al., 2009) or results from training studies that can elicit adaptations in muscle strength in addition to muscular coordination (Lee & McGill, 2017; Tayashiki et al., 2016). This confounds the interpretation of how trunk muscle coordination can impact the force exerted at the end-effector in multi-joint tasks. Furthermore, scarce evidence is available on how trunk muscle coordination should be cued to influence physical performance. Yet, trunk muscle coordination is cued in research and applied settings using a range of approaches from simple directives (Bressel et al., 2012; Richardson et

al., 1992; Tayashiki et al., 2015, 2016) to detailed coaching involving a series of verbal and physical cues (McGill, 2016; Starrett & Cordoza, 2015).

There is a scarcity of knowledge on how posture and trunk muscle coordination can influence physical performance such as the ability to exert force at the end-effector in multi-joint tasks (e.g. exert force at the hands during lifting). This has limited the development of quantitative approaches to assess how an individual's movement can influence their ability to exert force at the end-effector as well as strategies to modify movement to enhance physical performance. Investigating the effects of posture and trunk muscle coordination on physical performance can inform the development of more comprehensive strategies to assess and modify movement that considers both biomechanical demand in concert with physical performance. This can advance strategies to effectively regulate biomechanical demand without hindering physical performance as well as enhance physical performance without imposing unnecessary biomechanical demands.

1.1 Global Thesis Aim

The global aim of this thesis was to investigate how modifying posture and trunk muscle coordination can influence the ability to exert force during isometric lifting.

To address this global aim, 4 studies were performed. The first 2 studies (Chapters 4 and 5) involved within participant comparisons of isometric lifting strength across different postures and trunk muscle coordination patterns, to investigate how modifying these movement features may impact the ability to exert force at the hands in a position emulating lifting from the ground. Study 3 (Chapter 6) examined if existing kinematic-

based proxies for leverage could statistically explain the individual variation in the effects of posture on isometric lifting strength that were observed in Study 1 (Chapter 4). Given the heterogeneity observed in the effects of posture on isometric lifting strength observed in Study 1 (Chapter 4), Study 4 (Chapter 7) investigated if individuals self-organize their whole-body posture to maintain their ability to exert isometric lifting force, in response to cues targeting low back posture.

1.2 Specific Research Questions, Objectives and Hypotheses:

The specific research questions and objectives are summarized in Figure 1.1. What follows is a brief elaboration of the rationale supporting each research question, objective, and hypothesis.

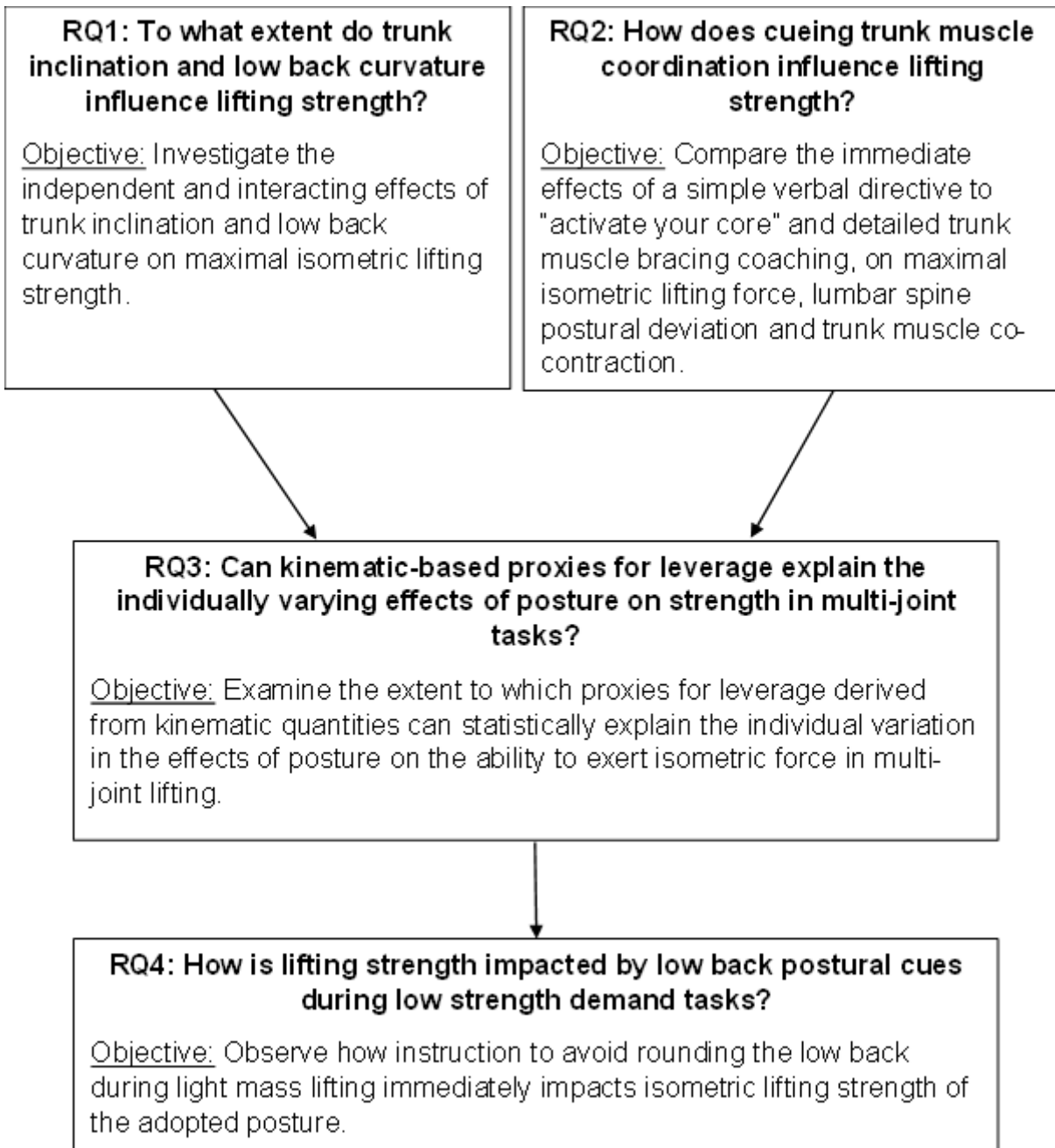


Figure 1.1: Overview of thesis research questions (RQ) and objectives. Research questions 1 to 4 represent Studies 1 to 4.

Research Question 1: To what extent do trunk inclination and low back curvature influence lifting strength?

Study 1 Rationale: Trunk inclination and low back curvature have long been targeted in movement assessment and coaching within both health and human performance settings, as they uniquely influence low back biomechanical demand. Each of these postural characteristics can also be independently modified. Trunk inclination can be modified by whole-body postural changes involving the ankles, knees, and hips without requiring changes in low back curvature, whereas low back curvature can be modified through pelvis and ribcage rotation. Modifying these postural characteristics may also distinctly influence the ability to exert force at the end-effector in multi-joint tasks, which can influence task performance. However, the effects of modifying trunk inclination and low back curvature on force exertion at the end-effector during multi-joint tasks have not been dissociated.

Study 1 Objective: Investigate the independent and interacting effects of trunk inclination and low back curvature on maximal isometric lifting strength.

Study 1 Hypothesis: It was hypothesized that the effects of changing trunk inclination on maximal exerted lifting force would be greater than changing low back curvature since changing trunk inclination would influence the contributions of more joints and muscles to the force exerted at the end-effector (i.e. hands). It was also hypothesized that the effects of trunk inclination and low back curvature would interact such that the effect of either postural characteristic would depend on the other.

Research Question 2: How does cueing trunk muscle coordination influence lifting strength?

Study 2 Rationale: Various approaches have been used in research and applied settings to cue trunk muscle co-contraction, which is colloquially referred to as bracing, ranging from simple verbal directives to detailed coaching involving combinations of verbal and physical cues. Cueing bracing has been shown to prevent low back postural displacement which can aid in increasing structural and tissue tolerance of the spine. Although co-contracting the trunk muscles can reduce the net effect of the agonist muscles, some evidence has suggested that bracing may potentially improve force exerted at the end-effector in multi-joint tasks. However, this previous evidence has not isolated if the observed effects are solely due to trunk muscle coordination and if cueing bracing is effective at immediately increasing force exertion in multi-joint tasks. Further, it is unknown if simple verbal directives are sufficient in comparison to detailed coaching of trunk muscle co-contraction in influencing muscle coordination and the ability to exert force at the end-effector.

Study 2 Objective: Compare the immediate effects of a simple verbal directive to “activate your core” and detailed trunk muscle bracing coaching, on maximal isometric lifting force, low back postural displacement and trunk muscle co-contraction.

Study 2 Hypothesis: It was hypothesized that detailed coaching would be more effective than simple directives at creating co-contraction that stiffens the spine to

reduce low back postural displacement and increase maximal end-effector isometric lifting force production.

Research Question 3: Can kinematic-based proxies for leverage explain the individually varying effects of posture on strength in multi-joint tasks?

Study 3 Rationale: Prior research has proposed kinematic quantities such as the joint-to-external force moment arms and predicted joint angle-dependent torque-generating capacities to infer leverage and hence the effects of posture on strength. If these proxies or predictors of leverage that are derived from kinematic quantities can reasonably capture the individual variation in the effect of posture on kinetic quantities like strength, it may then become feasible to make inferences about the effects of posture on strength in practice settings as there are kinematic measurement techniques that are simple to use and accessible. However, these kinematic-based proxies for leverage do not incorporate biomechanical properties known to affect multi-joint kinetics and the extent to which they may explain the individually varying effects of posture on the potential to exert force has not yet been verified.

Study 3 Objective: Examine the extent to which proxies for leverage derived from kinematic quantities can statistically explain the individual variation in the effects of posture on the ability to exert isometric force in multi-joint lifting.

Study 3 Hypothesis: Consistent with fundamental biomechanical principles, it was hypothesized that the variables derived from both kinematic and kinetic data (net joint reaction moments calculated using a rigid link model and inverse dynamics)

would explain a higher proportion of the individually varying effects of posture on the ability to exert isometric lifting force compared to the variables derived solely from kinematic quantities (external moment arms and predicted joint torque capacity).

Research Question 4: How is lifting strength impacted by low back postural cues during low strength demand tasks?

Study 4 Rationale: Movement-based interventions to reduce biomechanical demand may be ineffective and unsustainable if they hinder physical performance. Targeting specific movement features (e.g. lumbar curvature) as opposed to a single general whole-body postural technique (e.g. squat lift) hypothetically affords performers with greater flexibility to regulate biomechanical demand without compromising performance potential. However, using simple verbal directives to cue specific movement features during tasks with low external demands may not acutely prompt individuals to prioritize physical performance as they self-organize, eventually making the intervention ineffective. For example, responding to low back postural cues in a manner that reduces the ability to exert force during light mass lifting that does not acutely require high force exertion at the end-effector, can reduce physical performance potential by initiating and accelerating fatigue due to the increase in relative effort required to perform the task. However, it is unknown if every individual maintains their ability to exert force as they self-organize in response to low back postural cues.

Study 4 Objective: Observe how instruction to avoid rounding the low back during light mass lifting immediately impacts isometric lifting strength of the adopted posture.

Study 4 Hypothesis: Based on the results from Study 1, it was hypothesized that not every individual would respond to simple verbal instruction during light mass lifting in a manner that maintains strength.

Chapter 2 : Literature Review

2.1 Low Back Disorders

Low back disorders (LBD) are the global leading cause of disability (Vos et al., 2016; Wu et al., 2020) and has been associated with an increased risk of mortality (Martin et al., 2020). The prevalence of LBD has increased by over 54% between 1990 and 2015 (Vos et al., 2016), creating enormous health care and worker disability costs (Hartvigsen et al., 2018). In this thesis, LBD are considered to broadly encompass injury or pain in the low back region.

The majority of LBD cases have been categorized as non-specific where a pathoanatomical cause cannot be determined (Han et al., 2022; Maher et al., 2017). Koes et al., (2006) suggested that over 90% of cases are considered non-specific. Only a few causes such as infection, compression fracture, spondyloarthropathy, malignancy, radiculopathy, and spinal canal stenosis have been accepted as specific causes (Maher et al., 2017). There is a vast diversity and complexity in individual patient presentation, influenced by comorbidities, biological, psychological and social factors, that can modulate pain processing (Hartvigsen et al., 2018). This multifactorial complexity has greatly challenged the ability to identify subgroups within non-specific LBD, but does not reduce the need for identifying biological mechanisms in phenotypically distinct subgroups (Buchbinder et al., 2018). In fact, the prominent belief among clinicians is that non-specific LBD are not a homogenous condition (Kent & Keating, 2004). Numerous clinical approaches have aimed to subgroup LBD in order to guide intervention (Hefford, 2008; Maluf et al., 2000; McGill, 2016; O'Sullivan, 2005). Treatment based on such

subgrouping approaches have also been shown to be more effective when compared to following general treatment guidelines (Fritz et al., 2003) but the magnitude of this effect in research to-date has been considered to be not clinically meaningful with no strong evidence supporting the selection of one subgrouping approach over another (Tagliaferri et al., 2022). Although many of these approaches can be considered evidence based, there is a limited ability to objectively quantify the causality inferred from the associated experimental evidence or assess the effectiveness of a specific approach which to some extent is obscured by the multifaceted complexity involved (Rabey et al., 2015). This has challenged the widespread adoption of any of these schemes (Chou et al., 2007; May et al., 2006).

Fundamentally, most subgrouping schemes that assist the targeting of clinical intervention involve a biomechanical based evaluation that aims to identify loading strategies and modify movement through non-pharmacological treatment, education and exercise (Karayannis et al., 2012). This is not surprising as biomechanics research has demonstrated a strong association between movement execution and LBD, attributed to the effect of movement execution on biomechanical demand (Butera et al., 2016; Corbett et al., 2019; Marras, 2008; McGill, 2016; Swain et al., 2020). There are complex biological, psychological and sociological factors that contribute to pain and can modulate pain processing (Hainline et al., 2017; Moseley, 2017). Although the movement biomechanics may not be deemed as the primary cause for a case of LBD when considering the multifactorial contributors, the importance of movement biomechanics and the need to improve our understanding of human movement through the broad and diverse lens of biomechanics is undisputable (Cholewicki et al., 2019). Most LBD certainly influence

movement, but perhaps more importantly changing movement can relieve pain from LBD (Cholewicki et al., 2019; Ikeda & McGill, 2012). In fact, the most effective known prevention and treatment for chronic LBD is movement in the form of exercise (Fernández-Rodríguez et al., 2022; Huang et al., 2020; Steffens et al., 2016). However, the prescription of movement training and exercise is greatly hindered by the current knowledge and understanding of mechanisms of action to inform targeted prescription (Wood et al., 2020). Even if the goal is to just maintain general fitness and avoid deconditioning (Tagliaferri et al., 2020), there is a need to execute the exercise in a way that does not create or exacerbate pain.

2.2 Lifting and Low back Disorders

There is evidence that lifting has been associated with LBD as early as the 1950s with recognition that the execution of the movement is of importance, and recommendations to use a squat lift technique (Bauwens & Coyer, 1955; Millikan, 1954). Since then, evidence has increased and supported the association between lifting and LBD reporting (Coenen et al., 2014; Nelson & Hughes, 2009; Norman et al., 1998; Wai et al., 2010). Evidence has also increased to elucidate mechanisms through which movement can influence internal tissue exposures which can contribute to the risk of experiencing LBD (Marras, 2008; McGill, 2016).

Lifting techniques such as the squat and stoop lift, have been investigated since the 1970s (Garg & Herrin, 1979) to understand the underlying biomechanical, psychophysical, psychological, and physiological factors that may support recommending a “proper” technique (Straker, 2003). Using advanced and sophisticated biomechanical models to estimate the internal tissue forces across these different lifting techniques has

not provided conclusive results (Bazrgari et al., 2007; Kingma et al., 2004, 2010; Potvin, McGill, et al., 1991; von Arx et al., 2021). This may be attributed to the differences in modelling procedures (von Arx et al., 2021) as well as their sensitivity to the inter-individual variation in body and spine posture that occurs in the execution of each technique (Abdoli-Eramaki et al., 2019; von Arx et al., 2021). The variation in the execution of each technique has also made it difficult to identify specific movement features that should be targeted to manage exposures. Sophisticated biomechanical models are difficult to replicate and not easily accessible leading most researchers to use the net joint moment to estimate and infer internal low back loading. Apart from the questionable effectiveness of promoting the squat technique to manage low back exposures (van Dieën et al., 1999), it continues to be focussed on as the best technique in manual materials handling training research and practice (Denis et al., 2020), as well as being perceived as the safest technique by both untrained individuals, trained individuals and trainers (Abdoli-Eramaki et al., 2019). Interestingly, adopting a squat technique can decrease strength (Kumar & Garand, 1992), increase energy expenditure (Garg & Herrin, 1979; Kumar, 1984; Welbergen et al., 1991), be perceived to be more exerting (Straker & Duncan, 2000) and reduce the ability to balance (Chow et al., 2005; Heiss et al., 2002). These factors can hinder task performance and have been suggested to be responsible for the poor adherence and adoption of the squat technique by individuals (Denis et al., 2020). Given these findings it is clear that movement behaviours can influence both exposure and performance.

To overcome the issues associated with conceptualizing movement as “techniques” and access to sophisticated approaches to estimating internal exposure,

some have recognized spine posture as a specific movement feature that can be targeted to manage exposures (Beach et al., 2018; Chan et al., 2022; Frost et al., 2015; Gagnon et al., 2016; Pinto et al., 2018). Spine flexion can increase the shear loading on the spine and redistribute forces to passive tissues (Khoddam-Khorasani et al., 2020; Potvin, McGill, et al., 1991; Zwambag & Brown, 2020). Repetitive and prolonged spine flexion can contribute to exposure but can also influence the viscoelastic state of tissues disturbing their function (Abboud et al., 2018; Pinto et al., 2021; Sánchez-Zuriaga et al., 2010; Solomonow, 2012). In addition to these changes in exposure, functional spinal units are also less tolerant in flexed postures (Callaghan & McGill, 2001; Gallagher et al., 2005; Gunning et al., 2001; Howarth & Callaghan, 2012) and less capable of actively counteracting potentially harmful anterior shear forces in flexed postures (Pinto et al., 2021; Potvin, Norman, et al., 1991). It is also relatively easier to estimate and interpret spine flexion compared to internal joint forces or the net joint moment which is not as easily relatable to internal joint forces. These factors have motivated targeting a reduction in spine flexion. However, targeting and evaluating spine posture alone may not be sufficient in modifying movement in an effective and sustainable manner. Given appropriate feedback, individuals may certainly decrease spine flexion but this accompanies varying changes in body posture and muscle activity which could be counteractive, inefficient or unsustainable (Pinto et al., 2018; Pinto & Dickerson, 2021). For instance, some may adopt a squat technique due to a preconceived notion that it is synonymous with avoiding spine flexion (Pinto et al., 2018), which may reduce spine flexion but may also hinder task performance and adherence to the intervention.

2.3 Effects of Movement on Pain and Injury

The way we move has been exclusively evaluated within the field of biomechanics, quantifying the movement execution and forces involved (Winter, 2009). External forces that act on the body from the external environment (e.g. gravity, handheld objects) as well as internal forces created by the body, expose the composite musculoskeletal tissues (e.g. muscles, ligaments, joints) and systems (e.g. circulatory and nervous system) to forces. This exposure can lead to injury or pain which is dependent not just on the instantaneous exposure, but the exposure history, properties of the tissue and pain sensitivity.

Tissues vary in composition and organization (i.e. structure) resulting in different mechanical characteristics such as strength and toughness. Biological tissues are also viscoelastic. Hence, their mechanical response can differ due to composition and hydration but are also dependent on the characteristics of the applied force such as magnitude, rate, duration, and direction (Kalkhoven et al., 2020; Kumar, 2001; Mueller & Maluf, 2002). Tissue architecture and non-uniformity make length or posture a modifier of its mechanical behaviour and tolerance (Gunning et al., 2001; Howarth & Callaghan, 2012; Kumar, 2001). Changes in mechanical behaviour can occur due to exposure history (e.g. frequency and duration of exposure causing fatigue or viscoelastic changes), which can reduce tissue tolerance (Kalkhoven et al., 2020; Kumar, 2001; Mueller & Maluf, 2002) and influence function (Pinto et al., 2021; Solomonow, 2012). Exposures exceeding the tolerance of a specific tissue causes damage (Kumar, 2001). Yet, due to the dynamic nature of tolerance low level exposures can be cumulative, resulting in reduced tolerance, eventually causing damage (Kumar, 2001).

When exposure exceeds the tolerance of a tissue, damage occurs which can provide a noxious stimulus to activate nociceptors to elicit pain (Hainline et al., 2017; Loeser & Treede, 2008). Sub tolerance exposures causing micro-damage, potential damage or inflammation may also provide a noxious stimulus (Hainline et al., 2017). The nervous system can also be influenced by exposures in a similar way to musculoskeletal tissues such that acute and repetitive exposures can cause disruption and damage, but can cause pain without provocation from a noxious stimulus, categorized as neuropathic pain (Hainline et al., 2017; Mueller & Maluf, 2002). In addition to these effects of exposure on the tissues of the body, the occurrence of pain is dependent on the sensitivity to pain, which can also be influenced by exposure to forces. Inflammation generated from exposures can decrease the pain threshold making an individual more sensitive to noxious stimuli and making innocuous stimuli painful (Marras et al., 2016). Physical exposures and their accumulation over time can also contribute to influencing psychological factors such as mood and anxiety as well as the nervous system which can influence pain sensitivity (Anish, 2005; Daenen et al., 2015; Hainline et al., 2017). Cumulative exposures can also impact lifestyle factors such as sleep that can inhibit recovery and increase pain sensitivity (Hainline et al., 2017).

Although exposure to forces can decrease both tissue tolerance and pain threshold in the short term, if appropriate and accompanied by recovery, adaptation can occur to increase tissue tolerance and pain threshold (Belavy et al., 2021; McGill, 2009). Yet, too low or too high of a stimulus can result in decreasing tissue tolerance and pain threshold (Daenen et al., 2015; McGill, 2009; Sluka et al., 2018). Adaptation can also be hindered without adequate recovery.

Given these multiple mechanisms through which exposures can influence the development and experience of pain and injury, it is imperative to manage exposures. Managing exposures may not only offer a means of risk reduction but also help enhance adaptation. Exposure management in physical training and ergonomics have focused on the task demands, regulating the external load, repetition, duration and frequency but also modifying or selecting tasks that require less mechanical work (Gabbett, 2016; McGill, 2009). However, these modifications are limited to the extent that the intended objective of the physical training or work needs to still be achieved. Since each type of tissue or system has different tolerances but also different capacities for recovery and adaptation (Buckwalter & Grodzinsky, 1999; Kannus et al., 2007; Vila Pouca et al., 2021), the internal distribution among these tissues is important to consider when trying to manage exposures (Kalkhoven et al., 2021).

The internal distribution of forces is primarily modulated by the way movements are coordinated and controlled (see section 2.3). If the internal distribution is transferred to tissues with lower tolerances or lower capacity for recovery and adaptation, the risk of these activities can be higher. Changes to a task may certainly reduce the occurrence of a high exposure but low exposures and exposure history are also important to consider due to the dynamic nature of tissue tolerance and pain sensitivity. Although changing a task can allow for reduced exposure and mechanical work required, individuals can vary in their movement behaviour in ways that counteract the benefits of any such changes (Faber et al., 2007). Similarly, efforts to increase tissue tolerance and pain threshold through prescribed movement (i.e. exercise) can be attenuated or even disadvantageous if the manner in which the movement is executed results in creating an inappropriate

stimulus or provides additional tolerance reducing exposure. This becomes even more vital when trying to increase tolerance and capacity, where progressively increasing exposure to stimulate adaptation is desired. The influence of movement can not only reduce the risk of developing pain and injury but may also help target the desired adaptation more effectively by distributing exposures to the tissues of interest and allowing for training more of the desired stimulus across the training cycle. Thus, exposure management should include managing how the movement is executed.

2.4 Evaluating the Effects of Movement on Exposure

Biomechanics research has advanced the estimation of the forces created and experienced by the body's composite musculoskeletal tissues and structures (e.g. muscles, ligaments, joints). These internal forces can be compared to the physiological, neurological and mechanical properties of the respective tissues and systems, to evaluate the impact of movement profiles (e.g. efficiency, control and exposure). Unlike external forces that act on the body from the external environment (e.g. gravity, handheld objects), internal forces are difficult to directly measure non-invasively. Biomechanical models provide a non-invasive means to estimate the internal forces which has greatly improved our understanding of how the body functions (Dreischarf et al., 2016).

To estimate internal forces, biomechanical modelling has created inverse and forward dynamics problems (Buchanan et al., 2004; Herzog, 1996). Inverse dynamics uses the applied external forces (e.g. ground reaction force), kinematics and anthropometrics as inputs into a model of the whole or a part of the body or limb as a group of linked rigid segments to determine the transfer of forces from the point of

application from the external environment, to the joint of interest (Hof, 1992). The net reaction moment about a joint of interest provides an estimate of the demand that the internal tissues of the joint need to meet. Using musculoskeletal models of the internal system, the net joint reaction moment can be distributed across the tissues to estimate exposure (McGill & Norman, 1986). Forward dynamics compares simulated outputs from theoretical musculoskeletal models to experimentally measured outputs, to determine how the system functions and the internal forces that occur during movement (Buchanan et al., 2004; Dowling, 1997). Both approaches offer unique insight into musculoskeletal function, hybrid approaches have been developed to help tune the process towards an estimate of internal forces that encompasses the unique features of each approach (Buchanan et al., 2004). Thus, there are a variety of approaches that have been used to estimate internal forces. In addition to variation in the general approaches mentioned, variation is introduced by differences in how the input data are collected and analyzed (De Luca, 1997; Della Croce et al., 2005; Edwards et al., 2011; Leardini et al., 2005; Winter, 1991), as well as the representation of anatomical features and their dynamics within a model (Barrett & Callaghan, 2021; Brown, 2004; Herzog, 1996; Zajac & Gordon, 1989). Each model has strengths and limitations, but unfortunately the outputs cannot be easily verified. Importantly, the development of these models has advanced the knowledge of what biomechanical factors contribute to the internal dynamics and how they do so.

Using these biomechanical models, the internal forces can be estimated during movement. These estimates can be compared to the generalized mechanical behaviour of the underlying tissues to interpret the impact of the movement on different tissues.

However, many of these models can be challenging to implement, especially outside of a biomechanics laboratory, as they require extensive data collection procedures and analyses as well as expertise, limiting their accessibility and utility. Additionally, these estimates only convey information of the resultant exposure, which is the compilation of multiple biomechanical variables sensitive to different movement features. In estimating the low back internal loading for instance, the net joint moment, body morphology and muscle forces can all contribute to internal loading (Dreischarf et al., 2016; Marras, 2012; McGill & Norman, 1986). The net joint reaction moment can be influenced by the position and velocity of body segments relative to each other, gravity and any external forces. The anatomical geometry will be influenced by spine posture and the muscle forces can be associated with neuromuscular activity. These movement features can differentially contribute to the estimated exposure but how they do so is not directly apparent from the exposure estimates. Thus, biomechanical models provide a tool to investigate the link between movement and exposure, but need to be used in a manner that elucidates this link.

2.5 Modifying Movement to Manage Exposure

Due to the influence of movement behaviour on tissue loading and tolerance, modifying movement can potentially assist in managing exposures (Beach et al., 2019; Cudlip et al., 2022; Faber et al., 2007; Frost et al., 2015; Ikeda & McGill, 2012; McGill, 2009; van Dieën et al., 1999). Managing the exposure of tissues to forces can help prevent overload and provide an opportunity for recovery and adaptation (Kalkhoven et al., 2020; Kumar, 2001; Mueller & Maluf, 2002). This is of particular interest in contexts such as repetitive work and physical training, as the chronic nature of these exposures can accumulate and

result in reducing tolerance and increasing risk of developing pain or injury (Hainline et al., 2017; Kalkhoven et al., 2020; Kumar, 2001; Mueller & Maluf, 2002). Yet, there is a need to understand what about movement should be targeted and how it should be targeted. Specifically, movement features that can be comprehended, observed, cued, and modified need to be identified (Knudson, 2021; McPherson, 1996).

In order to understand how different features of movement contribute to the resultant internal exposure estimates, the results from biomechanical modelling, typically joint forces, can be interpreted alongside the inputs by deconstructing or reverse engineering the result. This has helped understand how movement can influence exposure when comparing different groups or individuals such as novices and experts (Gagnon et al., 2016; Lett & McGill, 2006; McGill, McDermott, et al., 2009) or those with and without LBP (Marras et al., 2001). However, this approach only provides examples of how different movement features can modify exposure. It does not provide sufficient information to guide movement evaluation without estimating internal exposure nor does it entirely inform on how different movement features should be prioritized and targeted to manage exposures across different tasks and environments.

Towards creating evidence to inform how movement should be targeted to manage exposures, different types of movement strategies that can be used for a given task, have been categorized as “techniques” to allow for comparison. For example, a squat lifting technique that focuses primarily on bending the knees and keeping the torso upright, has been compared to a stoop lifting technique where the torso bends forward while keeping the knees nearly extended (Straker, 2003). The internal forces primarily evaluated are lumbar spine joint compression and shear forces, showing inconsistent results across

studies (Bazrgari et al., 2007; Kingma et al., 2004, 2010; Potvin, McGill, et al., 1991; von Arx et al., 2021). Differences between studies have been attributed to variations in the modelling approaches used. However, using a consistent modelling approach revealed that the internal exposure created by each technique can interact with the task constraints such that there is no ideal technique that can be recommended across every task (Kingma et al., 2004, 2010). Each context may require a modified or different technique, but it is impractical to empirically evaluate every situation to determine the best technique for every task. Further modification may be required to address inter-individual differences (Beach et al., 2019). Instructing participants to perform a squat or stoop technique can result in considerable variation in how the technique is performed, influencing the resultant exposure estimate (von Arx et al., 2021). This variation can be subject to an individual's interpretation (Abdoli-Eramaki et al., 2019) and capacity (e.g. joint mobility or strength) to perform the movement (Beach, Frost, Clark, et al., 2014; Li & Zhang, 2009; Patterson et al., 2022). Conceptualizing movement as a "technique" does not control the variation in orientation between body segments (i.e. body posture) or spine posture, each of which contribute distinctly to internal exposure (Dreischarf et al., 2016; Marras, 2012; McGill & Norman, 1986). The squat and stoop, for example, focus vaguely on whole-body position and does not control for differences in specific postural features such as spine posture which can influence the internal distribution of forces (Khoddam-Khorasani et al., 2020; Potvin, McGill, et al., 1991; von Arx et al., 2021). Thus, there is a need for an improved approach to target movement in exposure management, that encompasses how movement can be modified given the individual, task and environmental constraints (Denis et al., 2020).

Humans observe movement instantaneously as position in space and when considering the dimension of time, the motion can also be described, for instance by the speed at which the movement was performed (Skjaerven et al., 2008). Hence, it is feasible to qualitatively observe and comprehend kinematic movement features such as the relative position of different body segments with respect to each other (i.e. body posture) or the posture of a joint or set of joints such as the spine posture (Abbott et al., 2022). However, due to the different effects of body posture and spine posture on exposure, there is a need to distinguish between them. Muscle forces are not as easy to empirically measure or visually observe, but also distinctly influence exposure. This has led some to also cue and coach muscle coordination. Due to muscles having shorter moment arms relative to the spine compared to the moment arm of an external force, the muscles can be considered at a mechanical disadvantage, requiring them to generate much larger forces than the external force itself (Jorgensen et al., 2003). Additionally, muscles do not work in isolation but coordinate and co-activate in numerous different combinations, potentially requiring additional force production by agonist and synergistic muscles to account for antagonistic muscles (Granata & Marras, 1995). Although these trunk muscle forces impose compression on the spine, they are necessary for motor control (Gribble et al., 2003). The isolated osteoligamentous spine is a flexible column and buckles with approximately 90 N of compressive force (Crisco et al., 1992) requiring the trunk muscles to coordinate to stabilize the structure of the spine (Reeves et al., 2007). Hence, the low back has been conceptualized to require muscles to co-contract to provide stiffness due to the lack of structural stability from the skeleton and the multiple degrees of freedom available for motion. Researches have conceptualized a U-shaped

relationship describing a need for trunk muscle co-activity to provide stiffness to stabilize the spine, such that insufficient muscle activity and excessive muscle activity can both be problematic (Cholewicki & McGill, 1996). However, the extent to which muscle activity is insufficient or excessive is unknown and not defined. Due to the complex nature of muscle activity and coordination, it becomes very difficult to evaluate and target it to modify exposure.

Training the trunk muscles has been advocated as an important component to low back health as well as physical performance during daily living and athletics (Granacher et al., 2013; Huxel Bliven & Anderson, 2013; Saeterbakken et al., 2022). In addition to improving strength, endurance and tolerance, trunk muscle training is believed to improve muscular coordination and control of the spine leading to an enhanced ability to transfer forces and coordinate movement (Kibler et al., 2006). Practitioners have adopted different methods for cueing specific muscle activity and control, such as abdominal bracing to create trunk stiffness (Huxel Bliven & Anderson, 2013). Intentionally stiffening the spine through muscular co-contraction can provide stiffness and is hypothesized to increase stability (Grenier & McGill, 2007; McGill, Karpowicz, et al., 2009). Low back pain patients that lack sufficient stiffness may benefit from abdominal bracing (Hicks et al., 2005), and its use has carried over to exercise and physical training (Huxel Bliven & Anderson, 2013; McGill, Karpowicz, et al., 2009; McGill, McDermott, et al., 2009; McGill, 2011; McGill & Marshall, 2012). Abdominal bracing during light lifting, similar to what may be performed in the workplace, reduces lumbar spine motion and lumbar extensor activity (Coenen et al., 2017). Yet, it is unclear if adopting this activation technique sufficiently reduces exposure as the co-activity can increase joint forces (Grenier & McGill, 2007; McGill,

Karpowicz, et al., 2009) and cueing individuals to attend to their trunk muscle coordination can impair motor control (Reeves et al., 2006). There remains a need to understand how muscular activity should be evaluated and targeted beyond its effects on exposure.

2.6 Movement Effects on Physical Performance

Biomechanical movement evaluation has focused on evaluating exposures, typically at a specific joint of interest such as the low back. Consequently, movement has been targeted with the goal of reducing low back exposure. For instance, manual materials handling training has idealized and promoted the squat technique for lifting. Yet, its effectiveness has been questioned (Denis et al., 2020; van Dieën et al., 1999). The reduction in low back demand when performing a squat technique appears to be dependent on the task constraints that may influence the low back to external force moment arm (Kingma et al., 2004, 2010; van Dieën et al., 1999). When controlling for the distance between the object and the individual via reach distance, lifting strength is reduced using the squat technique compared to the stoop technique (Kumar & Garand, 1992). This reduction in strength can indicate a reduced mechanical advantage, explaining why the squat technique also requires higher energy expenditure (Garg & Herrin, 1979; Kumar, 1984; Welbergen et al., 1991) and is perceived to be more exerting compared to the stoop technique (Straker & Duncan, 2000). These factors can hinder performance and have been suggested to be responsible for the poor adherence to adopt the squat technique (Denis et al., 2020). Thus, it is important that changing movement to reduce exposure should not negatively influence performance.

Physical capacity refers to what an individual can do and is the sum of the output of multiple systems of the body (World Health Organization, 2007). It can be tested under standardized task conditions that isolate specific physical attributes or systems. An individual would need to utilize their capacity to accomplish tasks or activities. However, having the capacity to accomplish a task does not reflect task performance as it depends on how the task is executed (Lamb & Keene, 2017). Specifically, the way we move has the potential to influence the forces generated by the body and how they transfer to the external environment. An increase in the output force given an input force, which may occur due to better transfer of forces within the body to the external environment, indicating a mechanical advantage. However, an increase in the ability of the body to generate force may increase force exerted externally but presumably not the ratio of input to output force which generally defines the mechanical advantage (Salamon & Midha, 1998). Instead, this author proposes that the execution of movement can influence the generation, transfer and application of forces to the external environment which can be considered as affecting the biomechanical advantage. Accordingly, manipulating the execution of the movement to influence the generation, transfer or application of forces would modify the biomechanical advantage. This not only influences task performance at the maximum limit (e.g. strength) but also at submaximal levels by influencing efficiency (Biewener et al., 2004).

Movement of a joint changes the instantaneous length and velocity of the underlying muscles, influencing their ability to produce force (Heckman & Sandercock, 1996; Herzog, 1996). The transfer of forces through the musculotendinous unit is also influenced by length and velocity (Finni, 2006; Heckman & Sandercock, 1996; Zajac,

1989). Muscles impose forces on the segments to which they are connected, generating torques about the joints they cross. The mechanical advantage of each muscle on a joint depends on its moment arm which can change with muscular activity and joint position (Finni, 2006). Thus, the posture and motion of a given joint can influence the ability to apply force to the external environment. The force generated by the musculotendinous unit and the leverage provided by the muscle moment arm interact to create the overall moment of force about a joint. The joint position of maximal force generation can be different than the position with greatest leverage, which can vary across directions of joint motion and across joints (Frey-Law et al., 2012; Loren et al., 1996; Zatsiorsky, 2003).

Since the body consists of multiple linked segments and not just a single joint, the movement of the rest of the body also influences the ability to exert force. The position of adjacent joints can influence the contribution of bi-articular muscles (Kawakami et al., 1998). Muscles and motions at a given segment can influence other segments across the body (Zajac et al., 2002). For example, in the context of the spine, position of the hips has been shown to influence trunk extensor strength (Kocjan & Sarabon, 2014). Although efforts to reduce biomechanical demand typically focus on reducing demand at a single joint, such as the low back during lifting, observing competitive lifters demonstrated no relationship between the low back moment and lifting strength (Cholewicki et al., 1991). Additionally, trunk extensor strength alone does not predict lifting strength (Schenk et al., 2006), further suggesting that the externally applied force cannot be inferred from observing a single joint. The end-effector strength has been postulated to be dependent not just on the moments produced at each joint but their configuration, with the same generated moments producing different end point forces with different body postures

(Zatsiorsky, 2003). However, the extent to which changes in body posture and spine posture contribute to strength is unknown.

The coordination of muscle activity can also influence the ability to produce joint moments. Antagonistic muscle activity can work against the agonist muscles, but provides stability to the joint and controls movement (Gabriel et al., 2006; Gribble et al., 2003; McGill et al., 2003). Although excessive co-contraction increases joint loading (McGill, Karpowicz, et al., 2009), stiffening the spine may also influence the ability to transfer forces within the body and exert forces externally. Observing strength athletes has demonstrated that experts intentionally co-contraction to prevent spine motion during heavy lifting but to also enhance force transfer (McGill, McDermott, et al., 2009; McGill, 2011). In specific situations, the ability of the trunk musculature to create stiffness can also limit the ability to exert force distally at the hands or feet (Santana et al., 2007). Training the trunk musculature has been advocated for improving performance (Saeterbakken et al., 2022). Increased trunk flexor but not extensor strength, is associated with higher lifting strength (Schenk et al., 2006). Yet, muscular strength may not be as important as coordination when aiming to create trunk stiffness (Kawabata et al., 2010). Intentionally stiffening the trunk may improve the transfer of forces across the body but this remains to be thoroughly investigated.

Chapter 3 : Data Collection and Common Methodology

The data used for the studies in this thesis were collected within a single data collection session. Therefore, there is similar methodology with respect to equipment, calibration procedures and some data analysis procedures across the studies. This chapter describes the common methodology across studies. However, the experimental procedures (i.e. tasks), data analyses and statistical analyses will be described within each study chapter.

3.1 Participants

Forty participants (50% female; Male mean $\pm$ SD age 23 ± 3 years, mass 85.7 ± 13.6 kg, height 1.77 ± 0.07 m; Female mean $\pm$ SD age 23 ± 3 years, mass 66.3 ± 9.1 kg, height 1.69 ± 0.08 m) provided written informed consent prior to participation. Inclusion criteria for participation was to be between the ages of 18 to 64 and have no skin allergies or sensitivities to alcohol or adhesives. Exclusion criteria were any self-reported pain or injury within the last 6 months that prevented performance of daily activities, as this could potentially influence the ability to assume and exert maximal force in the targeted postural conditions. Participants reporting any other health condition required approval from a healthcare provider to participate in physically strenuous activities. The University of Waterloo research ethics board approved all study procedures.

To estimate the required sample size, a power analysis was conducted for a repeated measures ANOVA with a within-between interaction using G Power (Faul et al., 2007). A sample size of 20 was estimated to be able to detect a medium effect with 0.84 power. This was only used to approximate a minimum sample size needed as the linear

mixed effects modeling procedures that were originally proposed would need simulations and pilot data to estimate the minimum sample size, which were not available *a-priori*.

Eighty percent of the sample participants reported being physically active and having previous resistance training experience involving lifting.

3.2 Equipment

Kinematic marker clusters with at least 3 infrared emitting diodes were placed on participants' hands, upper arms, head, thighs, shanks, feet and on the trunk overlaying the seventh cervical, twelfth thoracic, and first sacral vertebrae (Optotrak Certus system, Northern Digital Inc., Waterloo, ON, Canada; Figure 3.1). Anatomical segment end points were digitized relative to the respective marker cluster to track each body segment (Table 3.1).

Table 3.1: List of digitized anatomical segment landmarks.

Anatomical Landmark Name	
1 st Metatarsal	Right and Left Aspect of the Head above the Ear
5 th Metatarsal	Acromion
Heel	Medial Epicondyle of the Humerus
Medial Epicondyle of the Femur	Lateral Epicondyle of the Humerus
Lateral Epicondyle of the Femur	Radial Styloid Process
Greater Trochanter	Ulnar Styloid Process
Posterior Superior Iliac Spine (PSIS)	2 nd Metacarpal
Anterior Superior Iliac Crest (ASIS)	5 th Metacarpal
Iliac Crest	

Lifting force was exerted against a handle (length: 40.1 cm; circumference: 10 cm) that was attached in series with a uniaxial load cell (MLP-1K, Transducer Techniques LLC, Temecula, CA, United States) using 2 carabiners, to a metal bar secured to the ground. The handle height was set to 39.5 cm to facilitate achieving the targeted postural conditions, as it was expected that not every participant would have the joint mobility to assume the desired postures using the height of an Olympic weight plate loaded barbel (22.5 cm). However, 4 participants required an increase in handle height by 6 cm to allow them to assume the desired postures for Study 1 (Chapter 4), which was kept consistent across all conditions for these participants and maintained for Study 2 (Chapter 4) to facilitate minimal low back flexion. The handle height was not changed for these participants for Study 4 (Chapter 7) as the procedures for this study were collected before Study 1 and 2 during the data collection session (See Section 3.3). The load cell was factory tested before the study and was calibrated at the start of each data collection session using up to 136.18 kg. Kinematic markers were placed on the handle and load cell to calculate the vector of the applied hand force (Figure 3.1 A).

Ground reaction forces were captured using a force plate under each foot (OR6, AMTI, Watertown, MA, USA; Figure 3.1 A).

Ag/AgCl surface electromyography (EMG) electrodes (Neuroline 720, Ambu, Ballerup, Denmark) were applied bipolarly and bilaterally on the latissimus dorsi, thoracic erector spinae, lumbar erector spinae, rectus abdominis, external oblique and internal oblique according to previously outlined procedures (McGill, 1991; McGill et al., 1996; Table 3.2; Figure 3.1 B). EMG signals were differentially amplified (AMT-8, Bortec

Biomedical, Calgary AB, bandwidth 10–1000 Hz; CMRR = 115 dB at 60 Hz; input impedance = 10 G Ω).

Table 3.2: Approximate locations for EMG placement.

Muscle	EMG Placement
Rectus Abdominis	$\approx$ 3 cm lateral to the umbilicus
External Oblique	$\approx$ 15 cm lateral to the umbilicus and at the transverse level of the umbilicus
Internal Oblique	below the external oblique electrodes and just superior to the inguinal ligament
Latissimus Dorsi	lateral to the T9 spinous process over the muscle belly
Thoracic Erector Spinae	$\approx$ 5 cm lateral to the T9 spinous process
Lumbar Erector Spinae	$\approx$ 3 cm lateral to the L3 spinous process

All data were captured synchronously using an ODAUII unit controlled by NDI First Principles software (Version 1.2.4, Northern Digital Inc., Waterloo, ON, Canada). Load cell, force plate, and EMG data were sampled at 2000 Hz while kinematic data from the infrared emitting diodes were sampled at 40 Hz.

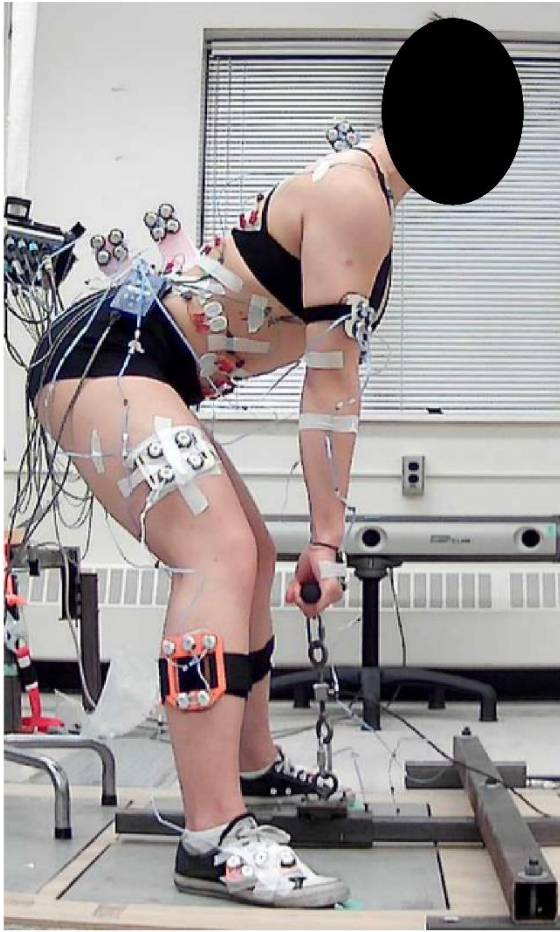
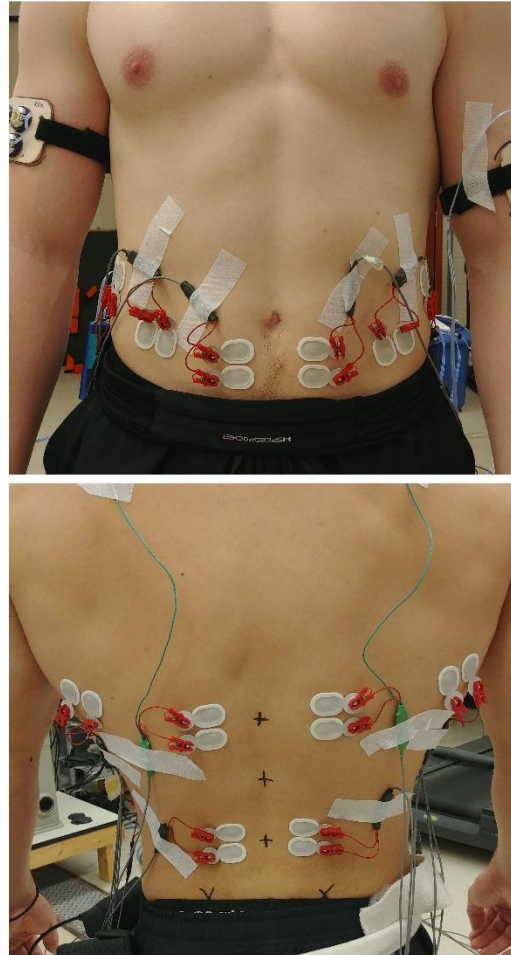
A**B**

Figure 3.1: (A) Experimental setup showing lifting handle and motion tracking marker placement during an isometric lifting exertion. (B) Trunk muscle EMG placement.

3.3 Procedures

An overview of the data collection timeline is provided in Figure 3.2 below. The main experimental procedures are within solid boxes and the reference trials within dashed lined boxes. Arrows show the flow across time.

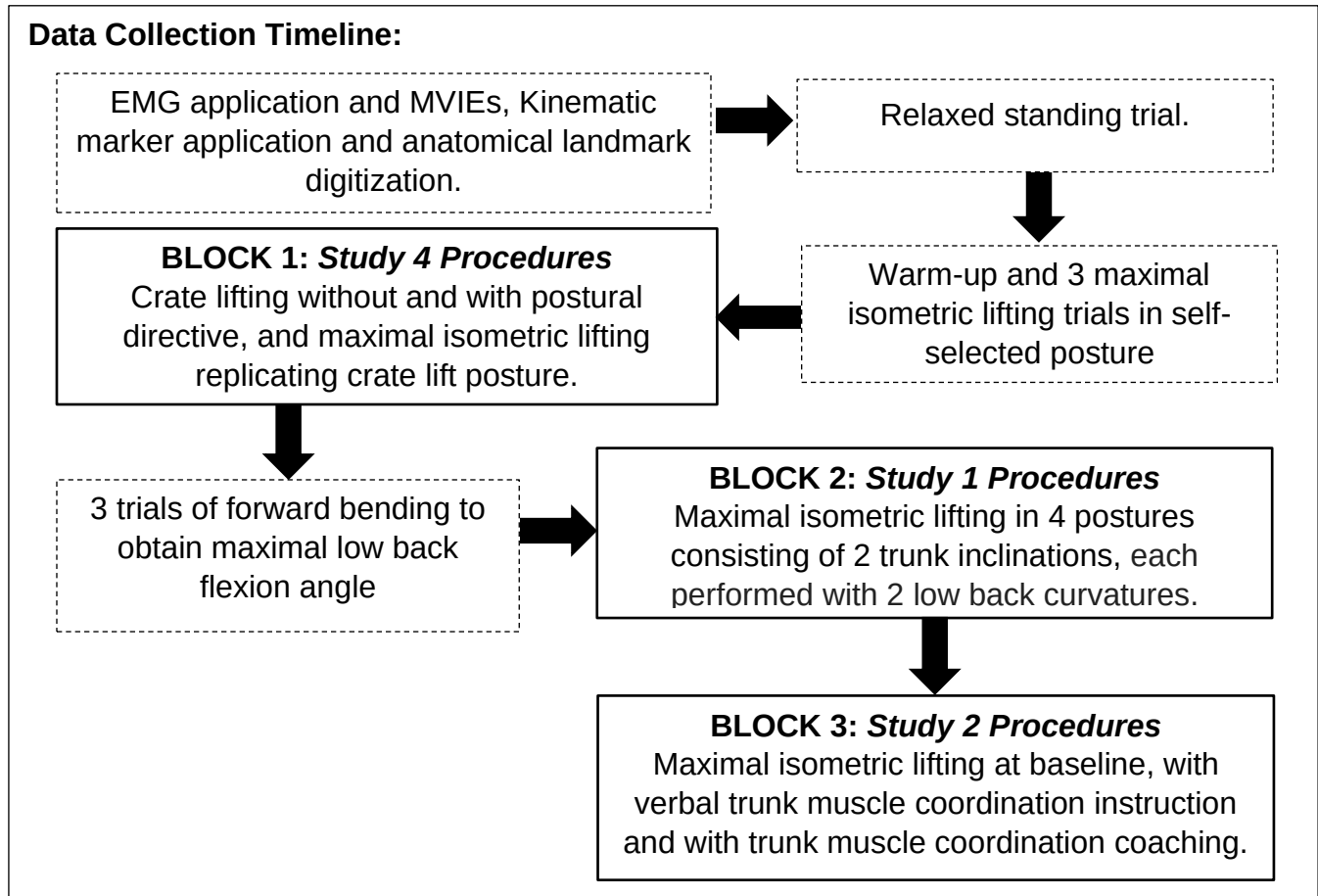


Figure 3.2: Entire data collection session timeline.

Study 1, 2 and 4 had independent experimental procedures that were collected within a single session. Each of these experimental blocks were conducted in the order indicated in Figure 3.2. Specifically, procedures for Study 4 (Chapter 7) were performed first, followed by procedures for Study 1 (Chapter 4) and Study 3 (Chapter 5).

Reference trials were collected before and after the first block of experimental procedures (i.e. procedures for Study 4). To normalize the EMG data, three trials of maximal voluntary isometric exertions (MVIE) for the trunk muscles were performed (Vera-Garcia et al., 2010). For the back muscles, participants laid prone with their legs

and hips supported by a massage table, and their upper body unsupported by the table. A researcher supported the participant's legs while another provided resistance at the upper back so that a back extension MVIE could be performed. For the abdominal muscles, participants sat at the edge of a massage table in a sit-up position, with their legs on the table supported by a researcher, while leaning their upper back on the second researcher. Abdominal MVIEs were performed by flexing, lateral bending to each side and twisting to each side against a matching resistance provided by the second researcher. This was followed by attaching the kinematic markers to the participant and digitizing the anatomical landmarks. Next, a 10-second relaxed standing trial was performed which provided data to generate a linked segment model of the body (Pinto & Callaghan, 2022). The joint angles in this standing posture were defined as representing 0°. Submaximal isometric lifting exertions were then performed as a warm-up and familiarization. Participants were instructed to apply efforts that they perceived to be approximately 30, 50 and 80% of their maximal effort. Following this, 3 maximal isometric lifting exertions were conducted in a self-selected posture.

After the procedures for the first experimental block (Study 4), 3 trials of forward bending were performed to obtain each individual's maximal low back flexion range of motion. Participants were coached on how to flex their low back by posteriorly rotating their pelvis and anteriorly rotating their ribcage. Starting from an upright standing position, participants maximally flexed their low back by rotating their pelvis and ribcage as they tried to bend their chest towards their hips. At volitional end range of low back flexion, a 3 second trial was collected during which they were verbally encouraged to further increase the low back flexion achieved. The maximum across these 3 trials was taken as

the maximum low back flexion angle that was used to normalize low back flexion angle. These trials were performed after the first experimental block to allow the researcher to provide cueing and feedback to achieve maximum low back flexion during these trials, without interfering with the natural responses to verbal directives that were investigated in Study 4.

Chapter 4 : The Effects of Trunk Inclination and Low Back Curvature on Lifting Strength

4.1 Introduction

Movement assessment and coaching across health (Beach, Frost, McGill, et al., 2014; Frost et al., 2015; Kent et al., 2015; McGill, 2004; Pinto et al., 2018) and human performance (Buckeridge et al., 2015; Cholewicki et al., 1991; Daly, 2001; Escamilla et al., 2000; Myer et al., 2014; Smith et al., 2008) foci have often targeted trunk inclination and low back curvature, as these postural characteristics have been linked to risk of low back injury. Trunk inclination and low back curvature can be modified independently from each other and are thought to have distinct effects on low back injury risk. Trunk inclination can be modified in multi-joint tasks such as lifting from the ground, by changing hip, knee, and ankle posture, without requiring changes in low back curvature (Carnegie et al., 2022). Modifying trunk inclination can influence the external forces and moments acting on the trunk which influences the overall biomechanical demand on the composite tissues of the trunk (Cholewicki et al., 1991; Escamilla et al., 2000; Hwang et al., 2009; Legg et al., 2017; Toussaint et al., 1992; van Dieën et al., 1999), but does not necessarily influence the distribution of forces across these tissues (Potvin, McGill, et al., 1991; Zwambag & Brown, 2020). In contrast, low back curvature can be modified largely through pelvis and ribcage rotation, and can modify the internal distribution of forces across the composite tissues of the lumbar spine region of the trunk (Potvin, McGill, et al., 1991; Zwambag & Brown, 2020). Specifically, a flexed low back curvature can migrate forces from active to passive tissues (Potvin, McGill, et al., 1991; Zwambag & Brown,

2020), which have reduced capacity and require longer durations for recovery and adaptation (Buckwalter & Grodzinsky, 1999; Kannus et al., 2007; Vila Pouca et al., 2021). Low back flexion can also reduce the tolerance of the functional spinal units to loading (Callaghan & McGill, 2001; Gallagher et al., 2005; Gunning et al., 2001; Howarth & Callaghan, 2012; Solomonow, 2012) while diminishing the ability of the lumbar extensor muscles to counteract potentially harmful anterior shear forces (McGill et al., 2000; Pinto et al., 2021). Thus, modifying trunk inclination or low back curvature can independently influence the biomechanical demands imposed on the low back and its capacity to withstand them. However, modifying trunk inclination and/or lumbar curvature to influence injury risk, may also influence task performance but in a manner that is not necessarily beneficial. In fact, the ineffectiveness of workplace training that promotes a vertical trunk inclination when lifting objects from the ground to reduce injury risk, has been attributed to an accompanying detriment to physical performance (Garg & Herrin, 1979; Kumar, 1984; Kumar & Garand, 1992; Straker & Duncan, 2000; Welbergen et al., 1991). Hence, considering the effects of trunk inclination and low back curvature on injury risk in conjunction with task performance, may inform the development of more comprehensive strategies to reduce injury risk and enhance physical performance. Yet, the effects of independently modifying trunk inclination and low back curvature on task performance has not been thoroughly investigated.

One way modifying trunk posture can influence performance in multi-joint tasks is by affecting the ability to exert end-effector force. For example, maintaining a vertical trunk inclination while lifting from the ground requires changing hip, knee, and ankle posture which influences the ability of each joint to generate torque and consequently

their contributions to the end-effector force. Specifically, modifying joint posture will influence the muscle-to-joint moments arms (Finni, 2006) and the length-dependent generation and transfer of muscle force across the musculotendinous unit (Finni, 2006; Heckman & Sandercock, 1996; Herzog, 1996; Zajac, 1989). Changes in posture in multi-joint tasks have also been hypothesized to influence the leverage of each joint on the end-effector force (Escamilla et al., 2000; Pinto & Dickerson, 2021; Swinton et al., 2011). In contrast, modifying low back curvature through pelvis and ribcage rotation, may primarily influence the force generated and transferred by muscles spanning the trunk and hips without considerably influencing muscles spanning adjacent joints such as the knees and ankles. However, it is possible that the effect of modifying low back curvature on end-effector force may depend on the trunk inclination. For instance, if changing whole-body posture to modify trunk inclination reduces the torque contributions from the trunk relative to the other joints, the effect of low back curvature on end-effector force may also be reduced. However, the independent and interacting effects of trunk inclination and low back curvature on end-effector force in multi-joint tasks have not been directly tested.

Prior research has not dissociated the effects of modifying trunk inclination and low back curvature on end-effector force production in multi-joint tasks. The primary approach used to investigate the effects of posture on multi-joint force exertion, has involved comparing task techniques that encompass a generalized whole-body postural profile rather than explicitly targeting posture of specific joints. For example, studies investigating the effects of trunk posture on lifting strength have compared techniques such as the squat and stoop lift (Kumar & Garand, 1992), or the conventional and sumo

deadlift (Cholewa et al., 2019; Cholewicki et al., 1991; Escamilla et al., 2000). The execution of these techniques can vary based on each individual's interpretation of the given instruction (Abdoli-Eramaki et al., 2019) and capacity to perform the movement (Beach, Frost, Clark, et al., 2014; Li & Zhang, 2009; Patterson et al., 2022), which results in individual differences in posture. Technique comparison studies have not typically accounted for individual differences in posture by statistically accounting or experimentally controlling for postural features such as trunk inclination and low back curvature independently, making it challenging to infer their independent and interactive effects on multi-joint force exertion. Although studies investigating the effect of trunk posture on isometric back extension strength while standing (Gallagher, 1997; Tan et al., 1993), sitting (Graves et al., 1990), kneeling (Gallagher, 1997) or side lying emulating standing (Daggfeldt & Thorstensson, 2003), show that bending the trunk forward can increase back extension strength, it is unclear if this is due to low back curvature or hip flexion (Kocjan & Sarabon, 2014), as these studies have not dissociated between trunk inclination and low back curvature.

A recent study investigated the effects of low back curvature on isometric back extension strength in a whole-body posture resembling lifting from the ground with a horizontal trunk inclination (Mawston et al., 2021). It was reported that a highly flexed low back curvature (98% of maximum low back flexion) allows for greater low back extensor torque generation during maximal isometric lifting exertions (Mawston et al., 2021). Apart from these results being based on a simplified estimate of the low back net joint reaction moment which may have been confounding, the implications drawn from these results promoted that a fully flexed low back curvature increases lifting strength. However, the

low back net joint reaction moment does not solely predict maximum mass lifted (Cholewicki et al., 1991), and the biomechanical contributions from each joint to the exerted force can be different when the hands are the end-effector as opposed to the upper back (Schellenberg et al., 2013). Importantly, promoting lifting with a fully flexed low back curvature may pose a higher risk of low back injury. Thus, there is a need to clearly understand the effects of modifying trunk inclination and low back curvature on the ability to exert end-effector force during multi-joint activities such as lifting as it will guide and advance movement-based assessments and interventions to manage injury risk and enhance performance.

The purpose of this study was to investigate the independent and interacting effects of trunk inclination and low back curvature on whole body isometric lifting strength. It was hypothesized that the effects of changing trunk inclination on maximal exerted lifting force would be greater than changing low back curvature since changing trunk inclination would influence the contributions of more joints and muscles to the end-effector force. It was also hypothesized that the effects of trunk inclination and low back curvature would interact. An isometric strength test was selected to facilitate control of trunk inclination and low back curvature when exerting maximal force. Lifting was selected as it has been associated with low back pain reporting (Coenen et al., 2014; Nelson & Hughes, 2009; Norman et al., 1998; Wai et al., 2010) which continues to be the global leading cause of disability (Vos et al., 2016; Wu et al., 2020) creating enormous health care and worker disability costs (Hartvigsen et al., 2018). To support a discussion of how modifying posture to increase strength can influence biomechanical demand, a secondary objective was to assess the effects of trunk inclination and low back curvature on lumbar

spine joint contact compression and shear forces. It was hypothesized that low back flexion would increase joint contact compression forces and anterior shear forces as low back flexion would reduce the oblique line of action of the extensor muscles to align with the compression axis of the spine (McGill et al., 2000; Pinto et al., 2021; Potvin, McGill, et al., 1991).

4.2 Methods

4.2.1 Experimental Procedures

Four lifting conditions were examined to dissociate trunk and lumbar spine contributions. These four conditions consisted of 2 trunk inclinations (vertical and horizontal), each performed with 2 low back curvatures (neutral and flexed, Figure 4.1). A researcher (BLP) first demonstrated and coached participants into the 4 postural conditions before performing any maximal exertions in the targeted postures. Participants were cued to find the most vertical and horizontal trunk inclination they could comfortably attain, by primarily bending their knees to keep the trunk vertical and primarily bending their hips posteriorly without letting their knees bend forward to maximize horizontal trunk inclination. Muscle activity was not cued to control low back curvature, to avoid interfering with the natural response.

Neutral and flexed low back curvature conditions were cued to be below 30% and over 80% of maximum low back flexion respectively. The threshold of 30% of maximum low back flexion was selected based on previous evidence showing that the lumbar spine vertebrae within the trunk are in neutral alignment with approximately 20° of low back flexion (Adams & Dolan, 1991) and that low back passive stiffness is minimized below

approximately 40% of maximum low back flexion (Beach et al., 2005; Fewster et al., 2021; Buchman-Pearle et al., 2023). Zero low back flexion (i.e. alignment documented during relaxed standing) was not targeted as the neutral curvature condition, as standing curvature can greatly vary between individuals (Zwambag et al., 2019). Individuals that stand in excessive low back extension would need to flex to align the spinal vertebrae. The flexed condition threshold of 80% of maximum low back flexion was selected as low back passive stiffness increases exponentially with the highest stiffness occurring past 80% of maximum flexion, indicating high strain on the passive tissues (Beach et al., 2005; Fewster et al., 2021).

Real-time optical motion tracking of low back, right knee, and right hip flexion/extension joint angles (NDI First Principles) were used during practice and subsequent experimental trials, to ensure the desired positions were achieved. Additionally, real-time video of the sagittal plane of the participant was displayed on a computer monitor in view of the participant to guide achieving the targeted postural conditions.

During the coaching period, participants were instructed to position their feet such that they could perform all 4 posture conditions without moving their feet. The position of each foot was restricted to two adjacent force plates each having a surface area of 0.23 m² (total surface area that stance was restricted to was approximately 0.5 m²). Once participants were able to assume each postural condition consistently, the placement of each foot was marked using tape before beginning the maximal isometric lifting trials. The handle was gripped with both hands pronated.

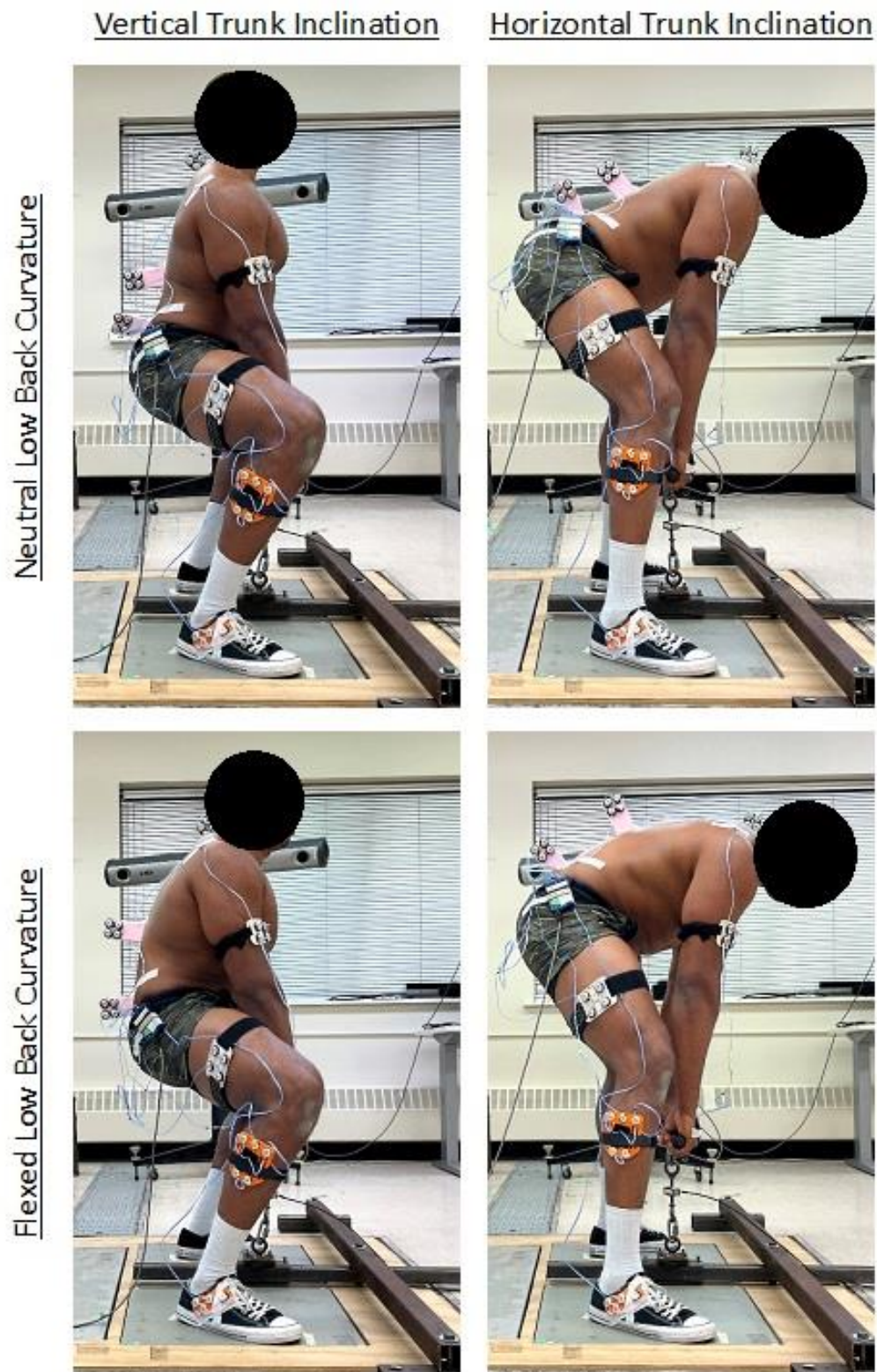


Figure 4.1: Depiction of the 4 posture conditions consisting of 2 trunk inclination (columns) and 2 low back curvatures (rows).

To facilitate consistency in trunk inclination and whole-body posture between curvature conditions within each inclination condition (vertical/horizontal), participants always assumed the respective trunk inclination with the neutral curvature during the setup for each trial, including the flexed low back curvature conditions. Once the initial setup posture was confirmed using real-time joint angle data, participants were instructed to flex their low back for the flexed curvature conditions, by posteriorly rotating their pelvis followed by anteriorly rotating their ribcage, in isolation, without moving other parts of the body. Visual observation and the real-time joint angle data were used to confirm motion was isolated to the low back.

Once the target posture was achieved and the participant confirmed they were ready, a 3 second countdown began followed by a command to start. Instruction was to keep the links that attached the handle to the ground taught by using as little force as possible during the countdown. Upon the start command, participants maximally exerted force at a self-selected rate. Instruction was to proceed as though they were lifting an object up from the ground. Verbal encouragement was provided by a single researcher spotting all participants, but it was first confirmed whether this encouragement motivated the participant to give a maximal effort. Participants were instructed to produce a maximum effort on each trial but were notified that they did not need to hold the maximum force for any duration of time and could stop once they were sure a maximum effort was provided, to prevent accumulation of fatigue.

Each postural condition was performed in a block randomized order with a minimum of 2 minutes of rest between each trial (Beckham et al., 2018; Chaffin, 1975; Kumar & Garand, 1992; Mital & Kumar, 1998).

4.2.2 Data Analysis

Kinematic data were processed in Visual 3D (Version 2022, HAS-Motion, Inc., Kingston, ON, Canada) and filtered with a dual pass Butterworth low pass filter with a 10 Hz cutoff (Kristianslund et al., 2012; Tomescu et al., 2018). Low back 3D joint angles were calculated relative to relaxed standing, using a flexion/extension, lateral bend and axial twist Cardan angle rotation sequence. Although prior research has calculated trunk inclination as the 3D orientation of the ribcage (Straub et al., 2021; Zwambag & Brown, 2015) or the 2D angle of a vector connecting the pelvis and ribcage (Faber et al., 2013; Khoddam-Khorasani et al., 2020; Melvin et al., 2010), modifying low back curvature through ribcage and pelvis rotation would impact these quantifications of trunk inclination. To limit this impact, trunk inclination was calculated as the 2D angle between the vertical axis of the environment and a vector connecting the mid points between the greater trochanters and acromia.

In Python (Spyder IDE, Version 5.3.0, Spyder Project Contributors), the vector between the markers on the handle and load cell was interpolated to 2000 Hz and used to find the vertical component of the maximum applied lifting force recorded by the load cell and the time instant it occurred. Trunk inclination and low back flexion/extension angle calculated from Visual 3D were extracted at the closest instant of kinematic data aligning or immediately prior (due to lower sample rate of the kinematic data) to the instant of peak force.

L4/L5 joint contact (i.e. bone on bone) compression and shear forces were quantified at the instant of peak isometric lifting force measured at the hands. A lumbar spine model that represents the resultant moment arm and line of action from an

anatomically detailed model with 50 muscle lines of action, as a single equivalent muscle, was used (Cholewicki et al., 1991). However, adjustments were made to make this single equivalent muscle model sensitive to how the geometry of the trunk muscles change with low back curvature to affect the moment arm and line of action of the resultant muscle force, providing a more biofidelic estimation of L4/L5 joint contact compression and shear compared to previous single equivalent models with a fixed moment arm and line of action (McGill & Norman, 1987; van Dieën & de Looze, 1999). The single equivalent for the L4/L5 reported by Cholewicki et al. (1991) has a moment arm of 6 cm and a line of action of 5° relative to the compression axis of the spine, which was assumed to be the orientation of the single equivalent muscle in each individual's respective relaxed standing low back posture (i.e. 0° of low back flexion). A quadratic polynomial equation reported by van Dieën & de Looze (1999) that estimates the low back curvature dependent moment arm of a single equivalent muscle relative to the L5/S1 joint, was modified to have a y-intercept of 6 cm to align with the L4/L5 model reported by Cholewicki et al. (1991). The goal of this analysis was to calculate joint loads at the level of L4/L5 instead of L5/S1 because spine loading at the L5/S1 is starkly different from the superior joint levels of the lumbar spine due to its orientation which results in very high joint reaction shear forces at the level of L5/S1 (Kingma et al., 2007). Additionally, the resultant effect of the muscles at the level of L5/S1 is additive to anterior shear force whereas the resultant effect of the muscles provides posterior shear forces at the superior joint levels of the lumbar spine (Kingma et al., 2007). The low back curvature dependent change in line of action of the single equivalent that was used to calculate L4/L5 joint contact forces in this study, was estimated by a linear equation based on previous research showing a

linear relationship (Pinto et al., 2021; van Dieën & de Looze, 1999). This linear equation was set to have a y-intercept of 5° representing the single equivalent muscle line of action in relaxed standing (i.e. when low back flexion is 0°) and an x-intercept at each individual's maximum low back flexion range-of-motion where the single equivalent muscle line of action was set to be 0° . The inputs to this single equivalent muscle model were the low back joint reaction forces, flexion/extension net joint reaction moment, and low back flexion/extension angle. The low back joint reaction forces and moments were calculated using a rigid linked segment model and bottom-up inverse dynamics using conventional Visual 3D (Version 2022, HAS-Motion, Inc., Kingston, ON, Canada) anthropometric data scaled to each individual's height and body mass. The joint centre of the low back was estimated as the mid-point between the iliac crests and the reaction forces and moments were resolved in the coordinate system of the thorax.

4.2.3 Statistical Analysis

A linear mixed effects model was used to account for the multilevel data structure and for potential heterogeneity in participant responses (lmerTest package, RStudio Version 2023.03.0, The R Foundation, Vienna, Austria). During data collection it was observed that participants were sometimes unable to maintain their setup position as they exerted maximal force. To account for differences in posture between individuals due to differences in mobility and loss of setup posture while exerting force, trunk inclination and low back flexion/extension angle were implemented as continuous variables in the model. This approach estimated the effects of posture on force as a regression slope for each individual, whereas implementing these conditions as factors would assume the postures were identical across participants. Accordingly, the full model consisted of the trunk

inclination, low back flexion angle and their interaction fitted as fixed effects as well as random effects with a random intercept and slope for each participant. Sex and order in which the postural conditions were performed were fitted as covariate fixed effects ($\alpha = 0.05$). A similar modelling approach was used to estimate the effects of trunk inclination and low back curvature on joint contact compression and shear forces, but order was not included as a covariate fixed effect as it was not expected to influence the effect of posture on joint loading.

Results from the linear mixed effects model showed high heterogeneity between participants, which motivated a *post hoc* examination of the individual differences to assess the extent to which the effect of modifying trunk inclination and low back curvature can be generalized. Participants were grouped based on individual changes in lifting strength in response to changing curvature within each inclination condition, and in response to changing inclination within each curvature condition. The standard deviation in maximum force across the 3 maximal isometric lifting trials which were performed using a self-selected posture at the start of the data collection, was multiplied by $\sqrt{2} \times 1.96$ to create a personalized confidence interval centered at 0 N (Swinton et al., 2018). This confidence interval approximated a threshold to categorize responses relative to each participant's ability to consistently produce maximum force. If the difference in the maximal lifting force between flexion and neutral curvature within each inclination condition, or between horizontal and vertical trunk inclination within each curvature condition, was less/greater than the confidence interval, the change was marked as a decrease/increase (D/I). If the difference in maximal lifting force was within the confidence

interval it was marked as no change (S). Only the maximum lifting force across the 3 trials from each condition was used for this *post hoc* analysis.

4.3 Results

Effects of modifying trunk inclination and low back curvature depended upon one another, as evidenced by a significant interaction between these postural characteristics in the linear mixed effects model (Table 4.1). However, there was high heterogeneity in responses to changing trunk inclination and low back curvature; the random effect standard deviation of the interaction, which indicates the variability of the effect between participants, was larger than the magnitude of the fixed effect estimate, which is the aggregate group effect. In addition to the co-dependence between trunk inclination and low back curvature, responses in lifting strength to modifying these postural characteristics were opposite to the aggregate trend estimated by the model for some individuals (Table 4.1).

Table 4.1: The linear mixed effects model results for the effect of trunk inclination and low back curvature on isometric lifting force. The estimates (β) for trunk inclination and low back curvature represents the change in lifting force (N) for every unit ($^{\circ}$) of horizontal rotation of the trunk or flexion of the low back whereas the interaction represents the change in force (N) for every unit of the product of trunk inclination and low back curvature. “” indicates a statistically significant effect.*

	<u>Fixed Effects</u>				<u>Random Effects</u>
	Estimate (β)	Standard Error	t value	p value	Standard Deviation
Intercept	1254.68	87.71	t=14.304	p<0.0001*	299.16
Trunk Inclination	-7.52	1.13	t=-6.655	p<0.0001*	0.72
Low Back Angle	-6.11	1.89	t=-3.233	p=0.0098*	4.98
Sex (Male)	455.99	82.51	t=5.527	p<0.0001*	-
Order	2.38	6.12	t=0.390	p= 0.6974	-
Interaction between Trunk Inclination and Low Back Angle	0.10	0.04	t= 2.624	p= 0.0119*	0.12

Table 4.2: Mean $\pm$ SD [range] in maximum lifting force, trunk inclination, and joint postures at the instant of peak force across the maximum trial for each participant, per condition. Flexion angle is represented as positive for all joints.

	<u>Vertical Trunk Inclination</u>		<u>Horizontal Trunk Inclination</u>	
	Neutral Curvature	Flexed Curvature	Neutral Curvature	Flexed Curvature
Maximum lift force (N)	1243.1 $\pm$ 515.8 [487.5-2333.3]	1191.8 $\pm$ 492.5 [360.7-2338.9]	1087.6 $\pm$ 421.4 [349.0-1782.6]	1112 $\pm$ 398.0 [472.6-2037.4]
Trunk Inclination	39.9 $\pm$ 5.9 [27.7-51.5]	36.4 $\pm$ 6.8 [23.0-53.6]	60.2 $\pm$ 6.1 [44.6-75.3]	58.5 $\pm$ 6.2 [37.8-70.2]
Low back flexion angle (°)	17.8 $\pm$ 7.3 [-0.3-35.4]	41.4 $\pm$ 8.7 [25.3-62.1]	20.7 $\pm$ 6.4 [0.8-40.3]	45.2 $\pm$ 8.4 [27.5-64.3]
Right hip flexion angle (°)	81.9 $\pm$ 8.4 [62.5-103.9]	60.6 $\pm$ 8.7 [42.7-79.5]	88.3 $\pm$ 7.0 [73.4-101.3]	63.1 $\pm$ 7.3 [46.1-77.9]
Right knee flexion angle (°)	70.7 $\pm$ 8.7 [53.5-89.7]	71.6 $\pm$ 8.6 [52.2-97.3]	41.1 $\pm$ 11.0 [17.8-69.7]	40.6 $\pm$ 10.5 [23.0-74.2]
Relative low back angle (% max flexion)	38.2 $\pm$ 14.5 [-0.6-70.1]	88.9 $\pm$ 11.2 [58.4-105.3]	44.6 $\pm$ 12.9 [2.0-76.8]	97.0 $\pm$ 7.6 [77.5-109.3]

Inclining the trunk horizontally decreased joint contact compression force but increased joint contact shear force whereas low back flexion increased both joint contact compression and shear forces as indicated by the regression slope estimates (Table 4.3). The impact of modifying low back curvature on joint loading was higher compared to modifying trunk inclination, especially for joint contact compression, as indicated by the regression slope estimates (Table 4.3) and aggregate data (Table 4.4). Although the

random effect standard deviation indicated heterogeneity in magnitude of the effect of trunk inclination and low back curvature on L4/L5 joint contact forces, they were smaller than the fixed effects regression slope estimates suggesting the direction (increase/decrease) of the effect of posture on joint loading was consistent across participants (Table 4.3).

Table 4.3: The linear mixed effects model results for the effect of trunk inclination and low back curvature on L4/L5 joint contact compression and shear forces. The estimates (β) for trunk inclination and low back curvature represents the change in joint contact force (N) for every unit ($^{\circ}$) of horizontal rotation of the trunk or flexion of the low back whereas the interaction represents the change in joint contact force (N) for every unit of the product of trunk inclination and low back curvature. “” indicates a statistically significant effect.*

	<u>Fixed Effects</u>				<u>Random Effects</u>
	Estimate (β)	Standard Error	t value	p value	Standard Deviation
<u>L4/L5 Joint Contact Compression Force</u>					
Intercept	5071.55	387.99	t=13.071	p<0.0001*	1343.62
Trunk Inclination	-28.02	6.12	t=-4.579	p<0.0001*	17.26
Low Back Angle	47.13	9.36	t=5.038	p<0.0001*	28.89
Sex (Male)	1434.97	301.83	t=4.754	p<0.0001*	-
Interaction between Trunk Inclination and Low Back Angle	-0.28	0.18	t=-1.526	p= 0.13	0.57
<u>L4/L5 Joint Contact Shear Force</u>					
Intercept	153.09	79.24	t=1.932	p=0.0691	243.73
Trunk Inclination	10.76	1.60	t=6.720	p<0.0001*	7.90
Low Back Angle	12.17	1.98	t=6.144	p<0.0001*	7.58
Sex (Male)	128.01	86.77	t=1.475	p=0.1431	-
Interaction between Trunk Inclination and Low Back Angle	-0.02	0.04	t= 2.624	p=0.6067	0.14

Table 4.4: Mean $\pm$ SD [range] in L4/L5 joint contact compression and shear forces across the maximum trial for each participant, per condition.

	<u>Vertical Trunk Inclination</u>		<u>Horizontal Trunk Inclination</u>	
	Neutral Curvature	Flexed Curvature	Neutral Curvature	Flexed Curvature
L4/L5 Joint Contact Compression Force	5640.2 $\pm$ 2126.2 [2873.3-10136.5]	6553.4 $\pm$ 2740.4 [3013.3-14797.3]	4877.3 $\pm$ 1585.1 [2409.7-8026.1]	5813.6 $\pm$ 1815.0 [3179.6-9505.9]
L4/L5 Joint Contact Shear Force	909.8 $\pm$ 363.5 [343.2-1773.9]	1132.3 $\pm$ 402.5 [390.5- 1986.0]	1129.8 $\pm$ 342.9 [611.9-1798.4]	1403.9 $\pm$ 411.8 [801.8-2273.9]

4.3.1 Post Hoc Investigation of Individual Differences

Seven different groups were identified in response to modifying curvature within each inclination condition, and 9 groups were identified in response to modifying inclination within each curvature condition. The curvature groups included those that showed the same response (increased, decreased or no change) (group II, DD and SS) and those that had different responses to low back flexion between the horizontal and vertical conditions (groups ID, DI, DS, SI where the first and second letter marks the response within vertical and horizontal inclinations respectively; Figure 4.2). The same groups were identified in response to changing trunk inclination from vertical to horizontal within each curvature condition, but with the addition of 2 groups: IS and SD (the first and second letter of the group marks the response within neutral and flexed curvatures respectively; Figure 4.3).

The largest increase and decrease in maximal exerted lifting force in response to low back flexion was 315 N and 620 N, and in response to inclining the trunk horizontally

was 296 N and 579 N, respectively. The changes that were marked as no change (S) ranged within -169 N to 145 N in response to low back flexion, and -324 N to 150 N in response to inclining the trunk horizontally, where the negative value represents a decrease in force with low back flexion or horizontal trunk inclination, respectively.

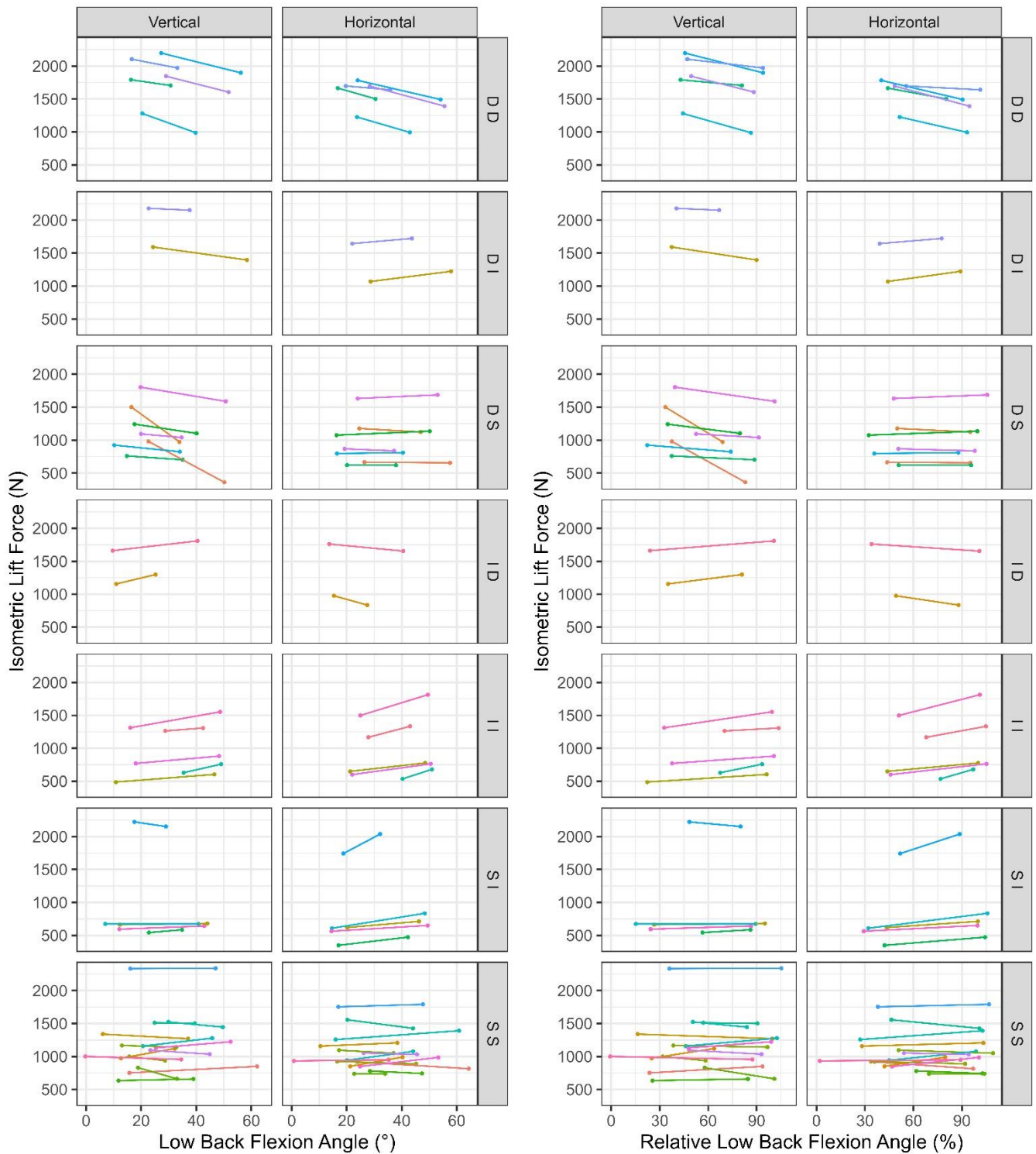


Figure 4.2: Participants demonstrated 7 types of responses to changing from neutral to flexed low back curvature across trunk inclination conditions. They increased (I), decreased (D) or did not change (S) their maximal exerted lifting force, similarly (groups II, DD, SS) or differently (groups ID, DI, DS, SI) across vertical and horizontal trunk inclination conditions. The first and second letter of the group name marks the response within vertical and horizontal inclinations respectively. Each participant is marked by a different colour. Responses are depicted relative to low back flexion angle in degrees (left) and normalized to maximum low back flexion angle (right).

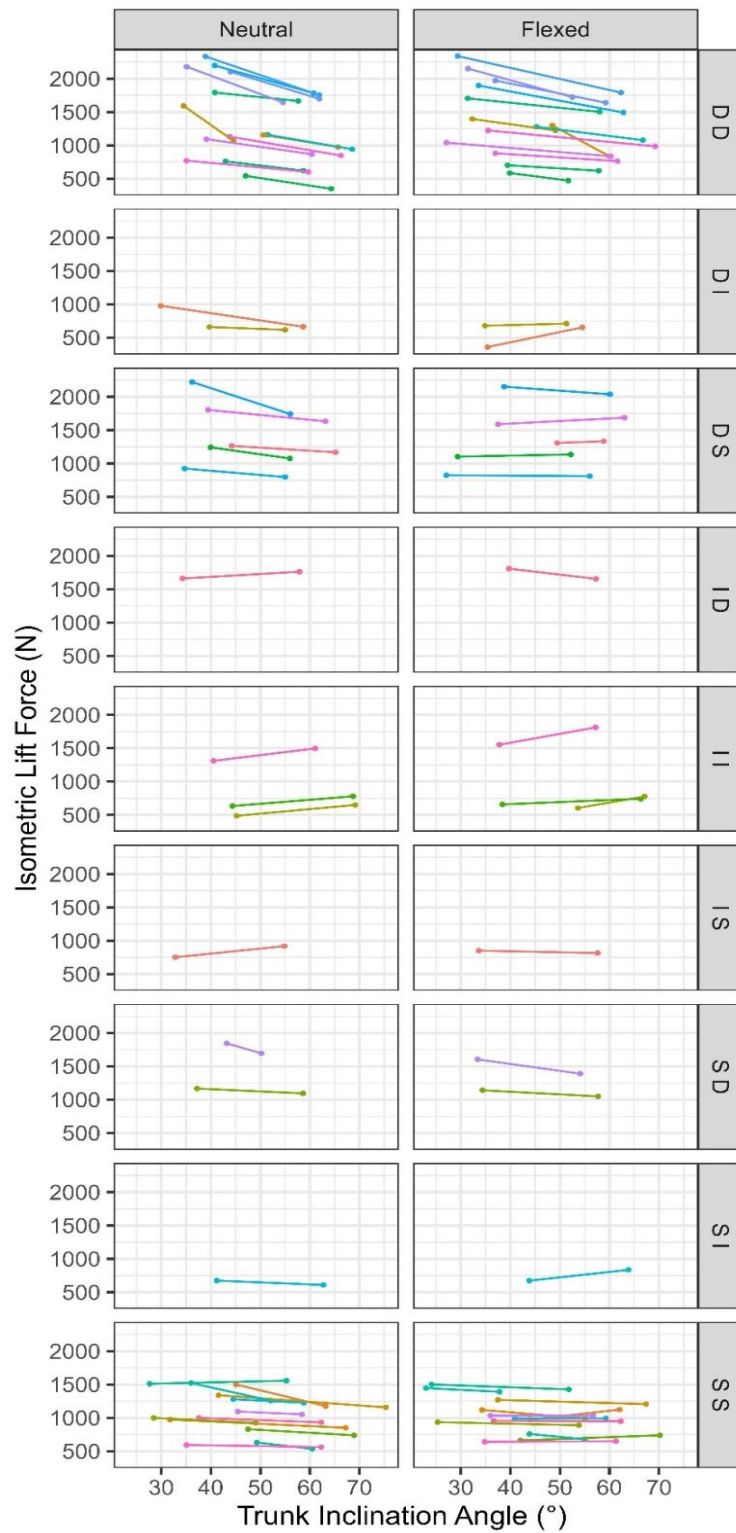


Figure 4.3: Participants demonstrated 9 types of responses in to changing from vertical to horizontal trunk inclination across low back curvature conditions. They increased (I), decreased (D) or did not change (S) their maximal exerted lifting force, similarly (groups II, DD, SS) or differently (groups ID, IS, DI, DS, SI, SD) across neutral and flexed low back curvature conditions. The first and second letter of the group name marks the response within neutral and flexed curvatures respectively. Each participant is marked by a different colour.

4.4 Discussion

Modifying trunk inclination and low back curvature each affected maximal isometric lifting force but did so inconsistently across participants. The first hypothesis of this study was rejected as the effects of modifying trunk inclination in comparison to low back curvature on the ability to exert lifting force was not greater. Both postural characteristics have a similar potential to effect isometric force exertion. The results confirm the second hypothesis that changing trunk inclination and low back curvature can interact to influence the ability to exert maximal end-effector lifting force. However, these interactive effects can greatly vary between individuals in magnitude and direction (increase/decrease). The high dimensionality of inter-individual differences suggests that the ability to exert force in multi-joint tasks cannot be solely inferred by observing posture and that a single postural profile cannot be generalized as being the strongest posture for all individuals.

The inter-individual variability in response to modifying trunk inclination or low back curvature on maximal isometric lifting strength, may be explained by individual differences that influence biomechanical factors. Differences in anthropometry such as segment lengths are associated with dynamic lifting strength differences between individuals (Cholewa et al., 2019; Ferland et al., 2020; Lockie et al., 2018) and hypothetically influence the mechanical advantage of the joint (Escamilla et al., 2000; Pinto & Dickerson, 2021; Swinton et al., 2011). However, the resultant effect on mechanical advantage may further depend on other factors such as range of motion. For example, a lack of hip (Patterson et al., 2022) or ankle (Beach, Frost, Clark, et al., 2014; Fuglsang et al., 2017) mobility may constrain the configuration of the kinematic chain, which may lead to less substantial changes in exerted force. Interestingly, the identified response groups within

the current study do not appear to be characterized by the trunk inclination or low back curvature range of motion exhibited by the participants (Figure 4.2 and 4.3). Mobility of lower extremity joints may also need to be considered as modifications in these joint postures may vary across individuals for the same change in trunk inclination due to characteristics such as segment length. Since results of the current study suggest that posture cannot solely be used to infer an individual's ability to exert force, it is of interest to identify which biomechanical variables explain the effects of posture on the ability to exert force in multi-joint tasks. This can inform the development of assessments and interventions to enhance task performance.

Inter-individual differences in maximal isometric lifting force in response to modifying posture may also be explained by inter-individual variation in joint strength. Some individuals have stronger knees relative to their back, and vice versa (Li & Zhang, 2009). Shifting torque-generating demands to an individual's weaker/stronger joint by changing lifting posture may explain the resultant decrease/increase observed. The joint angle-torque relationship also differs between individuals due to adaptations to activities and training (Alegre et al., 2014; Herzog et al., 1991; Noorkõiv et al., 2014). Individuals varied in the curvature they exhibited during self-selected maximal isometric lifting, which could potentially be related to their personal joint angle-torque relationship of their back extensors. Changes from the setup posture were observed in some participants during the execution of the maximal isometric lifting exertions, which may be related to individual variation in the joint posture dependent torque capacity and/or differences in strength between joints. Such differences in joint strength between joints and joint angle-torque relationship may have been prominent in the current sample, as the participants ranged

from those that do not resistance train to competitive powerlifters. Understanding the extent to which individual differences in joint strength influence the response to modifying posture can elucidate the potential for developing tools to assess the effects of posture on multi-joint force production. It may also identify if training can modify individual responses to facilitate the adoption of trunk inclinations or low back curvatures that limit tissue loading while increasing tissue tolerance.

The current study identified that a single postural profile cannot be generalized and recommended as being the strongest for every individual. Thus, conclusions drawn based on aggregated data about the effects of trunk inclination and low back curvature on strength in multi-joint tasks should be questioned. There is a need to understand if individuals require postural coaching to maximize force exertion and how to approach modifying posture if necessary. With regards to the former, it cannot be assured that every individual can or will self-select the posture that will maximize force exertion without coaching. Individuals also vary in their ability to modify posture even when given identical postural cues or training (Frost et al., 2015; Pinto et al., 2018). An inability to modify posture may potentially be linked to an inability to find the posture that maximizes force production. If cueing or training is required to modify posture to enhance force exertion, it is then necessary to investigate how this should be accomplished in tandem with preserving musculoskeletal health. Targets or approaches for modifying posture should consider the task objectives and the potential for injury risk due to the concurrent effects of posture on tissue loading and tolerance. For instance, cueing individuals who are stronger in low back flexion to train and perform high force exertion tasks with the low back fully flexed, may increase risk of injury if training loads are not effectively managed.

Alternatively, if the individually varying effects of posture on force exertion are primarily due to an individual's joint posture dependent torque-generating capacity (i.e. joint angle-torque relationship), modifying this capacity to increase torque in neutral low back postures can facilitate adopting a posture that reduces tissue loading and increases tissue tolerance without reducing strength. Thus, there is a need to understand why the effects of posture vary across individuals and how to most effectively modify posture to enhance force exertion while limiting injury risk.

To support the above discussion about the need to consider both physical performance and biomechanical demand when modifying posture, low back loading was quantified for the conditions investigated in this study (Table 4.3 and 4.4). Unlike the varying effects of posture on isometric lifting strength, the effects of modifying trunk inclination and low back curvature on L4/L5 joint contact forces were consistent in directionality (Table 4.3). From the estimated spine loads, it is evident that there is a higher penalty imposed by flexing the low back compared to inclining the trunk horizontally on L4/L5 joint contact compression forces (Table 4.3 and 4.4). Although the difference in the impact of trunk inclination and low back curvature on L4/L5 joint contact shear forces were more similar in magnitude, the highest joint contact shear forces occurred with a combination of horizontal trunk inclination and low back flexion (Table 4.4). An important consideration when assessing the joint loading penalty imposed is that in addition to increasing joint contact forces, low back flexion also reduces the tolerance of functional spinal units to both compression and shear forces (Callaghan & McGill, 2001; Gallagher et al., 2005; Gunning et al., 2001; Howarth & Callaghan, 2012). The spine loading results align with known mechanisms on the effects of low back flexion on

muscular geometry (McGill et al., 2000; Pinto et al., 2021; Potvin, McGill, et al., 1991). Low back flexion decreases the moment arm and line of action of the single equivalent muscle to become aligned with the joint's compression axis which increases muscular contributions to compressive joint contact forces and decreases the ability of the muscles to provide posterior shear forces to counteract the anterior joint reaction shear forces, resulting in higher joint contact shear forces. At the group level, flexing the low back appears to only contribute small gains in lifting force (e.g. 25 N or 2.2% mean difference between neutral and flexed curvature for the horizontal trunk inclination condition; Table 4.2) compared to the penalty imposed on L4/L5 joint contact compression (e.g. 936.3 N or 19.2% mean difference between neutral and flexed curvature for the horizontal trunk inclination condition; Table 4.4) and joint contact shear (e.g. 274.1 N or 24.3% mean difference between neutral and flexed curvature for the horizontal trunk inclination condition; Table 4.4). At the individual level, the participant that had the highest absolute increase in isometric lifting force by flexing the low back gained 315.2 N of isometric lifting strength (21.1% relative to neutral curvature) in the horizontal trunk inclination posture and 241.8 N (18.4% relative to neutral curvature) in the vertical trunk inclination posture. However, this increase in lifting force gained by flexing the low back came with a penalty on joint loading. Flexing the spine increased joint contact compression force by 1762.5 N (29.4% relative to neutral curvature) and 938.4 N (50.8% relative to neutral curvature) within the horizontal and vertical trunk inclination conditions respectively. Flexing the spine increased joint contact anterior shear force by 562.6 N (37.4% relative to neutral curvature) and 582.4 N (72.9% relative to neutral curvature) within the horizontal and vertical trunk inclination conditions respectively. These results suggest avoiding high

magnitudes of low back flexion during high demand activities such as exercise training to develop strength or repetitive activities such as work. Additionally, the variability in the effects of modifying posture on the ability to exert force at the end-effector suggests that the effect of posture on the ability to exert force in a multi-joint task is not fixed and that individuals can adapt with training to being stronger in neutral low back postures to reduce tissue loading and increase tissue tolerance. However, the impact of posture on biomechanical demand appears to be more consistent in directionality supporting that movement coaching should target postures that help regulate biomechanical demand.

A strength of this study is the linear mixed effects statistical analysis which included the inter-individual differences in each participant's strength and change in strength due to postural modifications. This allows for a more comprehensive and accurate analyses of the data compared to typically used analyses that do not account for heterogeneity (Detry & Ma, 2016; Van Dongen et al., 2004). Not considering variation between individuals may lead to erroneous conclusions and conflicting reports that are biased by the aggregate of the sampled data (Van Dongen et al., 2004).

While grouping individual responses provided insight into response heterogeneity, the confidence intervals used to classify the individual responses were estimated from the within-participant standard deviation across only 3 trials. This may have mischaracterized the threshold for detecting a change for some participants, resulting in some potentially meaningful responses being deemed not statistically different. For example, this study observed many immediate changes in strength in response to modifying posture, that exceeded changes expected in deadlift 1-repetition maximum following 6 weeks of training (approximately 70 to 190 N increases depending on training

experience (Androulakis-Korakakis et al., 2021; Colquhoun et al., 2018; Zweifel et al., 2017)). However, changes of this magnitude were categorized as no change (group “S” as explained in Section 4.2.3) in the current study due to the crude estimate of the within-participant standard deviations, resulting in 25% of participants having confidence interval thresholds between ± 157 and ± 422 N (median (Q2, Q3): ± 107 (± 53 , ± 157) N). Since participants ranged in resistance training experience from untrained to competitive powerlifters, more trials may have been required to familiarize some participants with performing maximal isometric exertions. This would have allowed for improved estimates of the within-participant variation and categorization of individual responses. Hence, some practically meaningful changes may have been masked by the statistical approach in the current study and the number of individuals categorized as not substantially changing (S) does not suggest that modifying posture has nominal influence on force exertion in multi-joint tasks.

Although this study used real-time 3D optical motion tracking feedback to control posture, some participants were not able to maintain the setup posture as they produced maximal force. Hence, differences between participants in the range of postures adopted (Figures 4.2 and 4.3) may not only have been due to differences in joint mobility, but also due to participants losing the setup position as they produced maximal lifting force. It also appears that ability to maintain position varied between postural conditions within an individual, which may have also been related to biomechanical factors impacting their ability to exert force such as leverage, mobility, and joint posture dependent torque capacity. Participants may have moved into joint postures that were stronger, or more comfortable if they were close to the end range of their joint mobility. Careful consideration

was taken to ensure participants were able to comfortably assume the desired setup postures without requiring a high effort.

Factors such as learning and fatigue may have restricted the ability to produce maximal force in within the 3 trials per condition. Some participants reported feeling that some postural conditions felt weak, awkward, or uncomfortable. These reported perceptions may have been influenced by a combination of many factors including the biomechanical factors discussed above. It is possible that some responses could have been different with additional practice. The data used in this study was part of a larger data collection which also included maximal voluntary contractions of the trunk muscles (Vera-Garcia et al., 2010), 4 maximal isometric lifting exertions and 13 dynamic lifts of a crate weighing approximately 10 % of the maximal isometric lifting force in the self-selected posture trials. Previous studies have investigated 56 (Kumar & Garand, 1992) and 90 (Chen et al., 2011) maximum isometric lifting exertions and 8 (Beckham et al., 2018), 12 (Guppy et al., 2019) and 18 (Comfort et al., 2015) maximal isometric mid-thigh pulls within a single data collection session, but did not report observing fatigue. Participants were frequently asked about their fatigue and although some reported being slightly fatigued compared to the start of the data collection, they asserted that they did not feel like it influenced their ability to exert a maximal effort within the postural conditions investigated here. This is supported by the lack of an effect of the order in which the conditions were performed (Table 4.1). Further research is required to determine the consistency of multiple maximal isometric lifting trials and the extent that learning and fatigue could have influenced the current findings.

4.5 Conclusion

Modifying both trunk inclination and low back curvature can influence ability to exert force in multi-joint tasks, but the effect can vary in magnitude and direction (increase/decrease) across individuals. This high variation in individual responses suggests that the ability to exert force cannot be inferred from posture alone. Contrary to previous notions drawn from group level analyses, a single postural profile in a multi-joint movement cannot be generalized as being the strongest for all individuals. Although some individuals may exert higher isometric lifting force in postures that are unfavourable for tissue loading and tolerance (e.g. low back flexion), the variation in the effect of posture on the ability to exert force suggests potential for individuals to adapt with training to be stronger in postures that are favourable for tissue loading and tolerance. In fact, the estimated joint contact forces suggest that training individuals to be stronger in more neutral ranges of spine posture should be the target for movement coaching due to the cost of flexing the low back on joint loading and tolerance.

Chapter 5 : The Effects of Cueing Trunk Muscle Coordination on Isometric Lifting Strength

5.1 Introduction

The isolated osteoligamentous spine buckles with approximately 90 N of compressive force (Crisco et al., 1992), requiring the trunk muscles to coordinate to control the structure of the spine and prevent buckling (Reeves et al., 2007). In comparison to targeting activity of specific muscles such as the transverse abdominus by cueing abdominal hollowing, cueing co-contraction of all trunk muscles which is colloquially referred to as bracing, has a higher potential to stiffen the structure of the spine by restricting intervertebral rotations to make it more robust to buckling (Grenier & McGill, 2007; Stanton & Kawchuk, 2008; Vera-Garcia et al., 2007). Stiffening the spine in an approximately neutral intervertebral alignment may further increase the load-bearing tolerance of the spine and reduce the risk of injury (Callaghan & McGill, 2001; Gallagher et al., 2005; Gunning et al., 2001; Parkinson & Callaghan, 2009). Immediate reductions in low back postural displacement which is a gross estimate of changes in lumbar spine intervertebral alignment, have been elicited within a single session by cueing individuals to brace during low force demand activities such as a single leg raise (Liebenson et al., 2009), sudden loading using weights less than 7 kg weights (Norrie & Brown, 2020; Vera-Garcia et al., 2006, 2007) and lifting 15 kg from the floor (Coenen et al., 2017). However, it is unknown if cueing bracing can immediately reduce low back postural displacement during high force or maximal demand tasks.

Interestingly, cueing bracing has also been suggested to improve force exertion at the end-effector in multi-joint tasks. Elite strength athletes have been observed to co-contract their trunk muscles when handling heavy loads (McGill, McDermott, et al., 2009). Additionally, training trunk muscle bracing for more than 6 weeks can improve impact force during martial arts striking maneuvers (Lee & McGill, 2017) and increase force production during isometric trunk and hip extension, as well as maximum velocity during a (dynamic) barbell squat (Tayashiki et al., 2016). Since these previous investigations have studied the resultant effects of cueing co-contraction over weeks or longer, it remains unclear whether the observed improvements in force production are solely due to cueing trunk muscle coordination or due to the physiological adaptations such as increases in muscle strength, that can occur over time in response to the cues and training. Although some co-contraction is necessary for spine motor control (Gribble et al., 2003; McGill et al., 2003), cueing co-contraction of the abdominal muscles has the potential to reduce the net effect of the agonist muscle moments. Additionally, bracing to stiffen the spine and avoid postural displacement may theoretically enhance force production during a multi-joint task by maintaining the link length of the trunk and its muscles, and reducing energy dissipation as forces are transferred across the spine (McGill, McDermott, et al., 2009; McGill et al., 2010; McGill, 2011; Santana et al., 2007). However, there is a scarcity of direct evidence to confirm if cueing individuals to brace and stiffen their spine will immediately enhance force exertion in multi-joint tasks. Investigating the immediate effects of cueing trunk muscle bracing may elucidate the extent that trunk muscle co-contraction influences the ability to exert force at the end-effector in multi-joint tasks, without the confounding effects of adaptation to training. This

can inform on how muscle coordination can influence an individual's physical performance potential and advance strategies to assess the effects of movement execution on physical performance. Additionally, it can inform if trunk muscle coordination should be cued to acutely influence physical performance.

Scarce evidence is available on how to cue bracing and whether it is even necessary to explicitly coach individuals on how to coordinate their muscles to create co-contraction and spine stiffness. Many studies investigating bracing, simply state that they instructed participants to co-contrast (Bressel et al., 2012; Richardson et al., 1992; Tayashiki et al., 2015, 2016). This simple verbal directive may be similar to the cue “activate your core”, which is used in professional practice contexts (e.g. fitness training, musculoskeletal rehabilitation, job coaching, etc.). Yet, a simple verbal directive may not elicit consistent outcomes across individuals as the interpretation and implementation of the directive can vary due to the accompanying ambiguity. Some studies have cued bracing through real time biofeedback using electromyography (EMG) over the external or internal oblique muscles (Brown et al., 2006; Grenier & McGill, 2007; Vera-Garcia et al., 2006, 2007). Although hollowing and bracing muscular coordination strategies have different muscle activation targets and motor execution, cueing these distinct muscle coordination strategies can similarly increase both external and internal oblique EMG (Vera-Garcia et al., 2007). Thus, EMG biofeedback of the abdominal oblique muscles cannot solely cue distinct muscle coordination profiles, suggesting that specific coaching may be required. Some studies provide physical cues by perturbing an individual's lumbar spine posture and instructing them to resist low back postural displacement through co-contraction (Liebenson et al., 2009), or by palpating and prodding the external oblique

muscles to provide a specific target to contract against (McGill & Karpowicz, 2009; Norrie & Brown, 2020). In the practice of movement training and coaching, similar coaching techniques which include physical cues along with more detailed verbal instruction have been popularized for teaching bracing which target spine posture, muscle contraction, and the Valsalva manoeuvre to create intra-abdominal pressure within the context of high force demand activities such as strength training (McGill, 2016; Starrett & Cordoza, 2015). However, it is unknown if there is a difference between cueing bracing using simple verbal directives or detailed coaching cues such as those used in applied settings.

The objective of this investigation was to compare the immediate effects of a simple verbal directive to “activate your core” and detailed trunk muscle bracing coaching, on maximal isometric lifting force production, low back postural displacement and trunk muscle co-contraction. Peak vertical isometric lifting force, low back postural displacement and trunk muscle co-contraction were evaluated across 3 conditions: baseline; directive; and coached. It was hypothesized that detailed coaching would be more effective than simple directives at creating co-contraction that stiffens the spine to reduce low back postural displacement and increase maximal isometric force production.

5.2 Methods

5.2.1 Experimental Procedures

Maximal lifting exertion trials consisted of 3 baseline trials, followed by 3 trials with the directive to “activate your core” and finally 3 trials using a coached bracing technique. Potential residual effects of coaching trunk muscle coordination demanded a sequential order of the experimental conditions so that the natural response could be observed

during baseline and directive conditions. For all trials, participants would first hold the handle and assume a setup lifting position which was based on their self-selected position during baseline trials. A start command was given following a 3 second countdown, after which participants began a ramp exertion until they were sure they exerted maximal effort. A minimum rest of 2 minutes was taken between each trial.

For the baseline trials, participants self-selected their lifting posture and stance which was restricted within the area of the 2 adjacent force plates (approximately 0.5 m²). Foot position was taped so that it could be maintained across subsequent trials. They were instructed to select a setup position that did not include excessive rounding or arching of their low back. The range in low back, right hip, and right knee flexion/extension joint angle during the 3 second countdown period across the 3 baseline trials, was used to guide participants into the same setup position for the subsequent directive and coached experimental conditions, using real time optical motion tracking feedback (NDI First Principles Version 1.2.4, Northern Digital Inc., Waterloo, ON, Canada). Three dimensional joint angles were calculated in NDI First Principles, as the orientation between diode rigid bodies on the relevant segments.

Before the performing the trials for the directive condition, participants were verbally instructed to “activate your core” while performing the maximal lifting exertion.

Prior to the coached condition, participants were coached through specific targets for intentionally bracing their trunk muscles, based on the work by McGill (2016) and Starrett & Cordoza (2015). Participants were requested to focus on what was specifically being coached, how it felt when the researcher confirmed that they successfully achieved the desired targets during coaching, and to create distinction between how they may

typically coordinate their trunk muscles so that they could avoid reverting to what they typically do. The coaching process began with a researcher using their fingers to prod both sides of the participant's lower trunk between the ribcage and pelvis, targeting the oblique muscles, while the participant was standing and relaxed. Participants were then asked to try to contract their muscles to push the researcher's fingers and resist the fingers from prodding into the abdominal cavity. Participants were informed that they should feel contraction all around their lower trunk and the researcher attempted to observe if the participant successfully created the desired contraction of stiffening the trunk without excessively protruding or hollowing the abdominals. Once achieved, the researcher once again prodded the participant's lower trunk as they were relaxed but gave the instruction to inhale in a way that caused the lower trunk to expand against the researcher's fingers. It was emphasized that this expansion due to inhalation was different from the previous muscle contraction. Once the participant was able to achieve the desired inhalation and expansion of the lower trunk, they were given approximately 1 minute to practice the muscle contraction and the inhalation without feedback from the researcher but were encouraged to repeat the prodding and steps that were performed during the researcher's coaching on their own. After this short practice, the researcher asked the participants to perform the muscle contraction and inhalation to ensure that they were able to successfully perform them. If needed, some additional feedback was given. The participant was then given a sequence of steps to brace their lower trunk. This included first maximally activating their trunk muscles as coached, followed by inhalation so that they felt pressure against their abdominal wall that they could contract against. The final step was to push the tip of their tongue against the bottom row of their teeth and

the middle of their tongue to the roof of their mouth. The aims of these steps were to first create trunk stiffness through muscle contraction, then increase intra-abdominal pressure with the coached inhalation, and finally use the tongue to create a Valsalva. Participants performed all three steps in sequence and tried to maintain the brace as they reached down to hold the handle, assume the lifting setup position, and then give a submaximal lifting effort. Participants were informed that they could perform a quick inhalation by “sipping the air” during the countdown proceeding the start command prior to beginning the maximal lifting exertion if they felt like they needed it, but to ensure that this breath did not change what the brace felt like during coaching or cause them to lose their brace. They were instructed to avoid exhaling but to focus on maintaining the brace, rather than trying to hold their breath, as it was assumed that focusing on holding the breath would distract participants from maintaining the brace. After the submaximal practice trial, participants were asked if they felt like they were able to successfully replicate the coached brace without the researcher’s help and could do so while performing a maximal lifting effort. Upon confirmation, 3 trials of maximal lifting exertions were performed using the coached bracing technique.

At the start of the data collection, participants were asked about their activity history as well as if they usually think about bracing or activating their core and how they cue themselves to do so. Towards the last trial of the coached condition, an informal unstructured interview was conducted where participants were primarily asked if they felt any differences between the 3 experimental conditions and if they perceived any differences in their ability to produce lifting force.

5.2.2 Data Analysis

A vector between the diodes on the handle and load cell was resampled to 2000 Hz and used to find the vertical component of the applied lifting force. The peak vertical lifting force and the instant it occurred was recorded. The onset of force production was identified using the first derivative method (Pinto & Callaghan, 2023). This included filtering vertical force-time signal using 10 Hz Butterworth dual pass low pass filter before calculating the first derivative of the signal. The onset was marked as the first instant of the longest time period during which the first derivative of the signal was continuously positive, between the first 1.5 seconds of the recording (i.e. middle of the 3 second countdown period) and the instant where the vertical force reached 50% of the maximum force exerted.

Motion tracking data were processed in Visual 3D (Version 2022, HAS-Motion, Inc., Kingston, ON, Canada). These data were first filtered using a dual pass low pass Butterworth filter with a 10 Hz cut-off (Kristianslund et al., 2012; Tomescu et al., 2018). Low back 3D joint angles were calculated relative to quiet standing, using a flexion/extension, lateral bend, and axial twist Cardan angle rotation sequence. Low back postural displacement was quantified as the difference in low back angle between the onset of the trial and the instant of time where peak lifting force was exerted. This difference in low back angle was normalized to each participant's maximum low back flexion angle to make the changes relevant to each individual's range of motion.

Co-contraction was estimated using an anatomically detailed EMG-driven musculoskeletal model of the lumbar spine, consisting of 118 muscle lines of action and 6 lumbar vertebrae each with 3 rotational degrees-of-freedom (Cholewicki & McGill, 1996;

Grenier & McGill, 2007; Howarth et al., 2008; McGill & Norman, 1986). This approach was selected to capture the combined effect of muscle activity, muscle force and anatomical orientation that influence the moments generated by the trunk muscles. As opposed to classifying muscles as being agonist and antagonist a priori, this approach estimated the contribution of each muscle line of action to the agonist or antagonist moment based on its posture dependent anatomical orientation. The model inputs were low back 3D joint angles and normalized linear enveloped EMG of the 12 trunk muscles collected in the current study.

EMG was linear enveloped, by first subtracting the bias (average across the signal), followed by full-wave rectification and finally filtering the EMG signal using a single pass second order low pass Butterworth filter with a 2.5 Hz cut-off frequency (Brereton & McGill, 1998). The linear enveloped EMG was then normalized to the peak linear enveloped EMG achieved during the MVIEs and down-sampled to 40 Hz to match the low back kinematic and kinetic data. EMG channels that appeared to have signal artifacts or loss of signal were replaced with the signal from the opposite side, assuming bilateral symmetry, before being input into the musculoskeletal model. This change was typically made across all trials within a participant to avoid influencing the comparisons between the experimental conditions and was applied at least to 1 EMG channel for 30 of the participants.

Within the musculoskeletal model, the vertebral and muscle kinematics were driven by the low back 3D joint angles. Activation of muscles that are too deep to observe using surface EMG were estimated from the collected surface EMG as follows: multifidus and quadratus lumborum from lower erector spinae, psoas and transverse abdominis

from internal oblique (Brown & Potvin, 2007; McGill et al., 1996). Muscle force was estimated using a Hill-type muscle model (McGill & Norman, 1986):

$$F_m = [NEMG_m \times PCSA_m \times \sigma_m \times l_m \times v_m + F_{pec}]$$

where F_m is the force produced by a muscle fascicle (m) about its line of action (N), $NEMG_m$ is the normalized linear enveloped EMG, $PCSA_m$ is the physiological cross-sectional area of each muscle (cm²), σ_m is the maximum stress that the muscle can potentially produce (set at 35 N/cm²), l_m is the length coefficient (unitless), v_m is the velocity coefficient (unitless), and F_{pec} is the force produced by the passive elastic component (N). The net flexor and extensor moments generated by muscles crossing the L4/L5 vertebral joint were estimated. Co-contraction was calculated as the proportion of flexor muscle moment relative to the extensor muscle moment. This approach considered for the resultant contributions of different muscle line of actions based on anatomical orientation and posture, to the flexor/extensor moments, rather than pre-assigning muscles as agonists or antagonists.

5.2.3 Statistical Analysis

Linear mixed effects models were used to investigate the within-participant changes across the 3 conditions (baseline, directive, coached) using the lmerTest package in RStudio (Version 2023.03.0, The R Foundation, Vienna, Austria). This statistical modelling approach allowed for the analysis to conserve all the recorded trials (3 trials x 3 conditions x 40 participants) by accounting for the multilevel data structure (i.e. multiple trials per participant) to avoid pseudo-replication. The linear mixed effects model also allowed for inclusion of heterogeneity in participant responses by estimating intercepts

and slopes for each participant. Data from 2 participants were excluded for the co-contraction analysis due to EMG artifacts that could not be accommodated by assuming bilateral symmetry.

A linear mixed effects model was created for each dependent variable (peak vertical lifting force, low back postural displacement, trunk muscle co-contraction). An additional model was created with the low back posture at onset of the isometric exertion as the dependent variable to assess if changes in low back postural displacement could be attributed to differences in initial posture. Each of these models included a fixed effect and random slope for condition as a factor with 3 levels (baseline, directive, coached), and a random intercept for each participant. Main effects of condition were examined from the model using a Type III Analysis of Variance with a Satterthwaite's method for estimating degrees of freedom. Significant main effects were further investigated using pairwise comparisons with a Tukey adjustment the multiple comparisons between conditions (emmeans package). Statistical significance was considered if $p < 0.05$.

5.3 Results

Peak vertical lifting force significantly decreased in the directive and coached condition compared to baseline (pairwise $p < 0.0001$), but the difference between the directive and coached condition was not significant (pairwise $p = 0.5671$; Figure 5.1, Table 5.1).

Compared to baseline, low back postural displacement slightly decreased during the directive condition, but this difference was non-significant (pairwise $p = 0.1690$), whereas displacement significantly decreased during the coached condition (pairwise $p = 0.0006$). The directive condition was not significantly different from the coached

condition (pairwise $p=0.1296$, Figure 5.2, Table 5.1). Low back posture at onset of the exertion was not significantly different across conditions (main effect $p=0.08605$, Table 5.1).

Trunk muscle co-contraction was significantly higher in the coached condition compared to baseline (pairwise $p=0.0003$) and the directive condition (pairwise $p=0.0071$). However, the directive condition was not significantly different from baseline (pairwise $p=0.3324$; Figure 5.3, Table 5.1).

Table 5.1: Estimated marginal means [95% Confidence Interval] for peak lifting force, low back postural displacement (change in low back posture from onset to the peak of the isometric exertion) and co-contraction for each condition.

	Baseline	Directive	Coached
Peak lifting force (N)	1194	1109	1096
	[1031, 1358]	[962, 1256]	[950, 1241]
Low back postural displacement	21.4	19.5	17.2
(% max flexion angle)	[17.6, 25.2]	[15.4, 23.6]	[13.2, 21.3]
Trunk muscle co-contraction (%)	10.1	11.0	13.5
	[8.58, 11.6]	[9.07, 12.9]	[11.08, 15.8]
Low back posture at onset of the	41.8	44.7	44.0
exertion (% max flexion angle)	[35.5, 48.1]	[38.0, 51.4]	[36.8, 51.1]

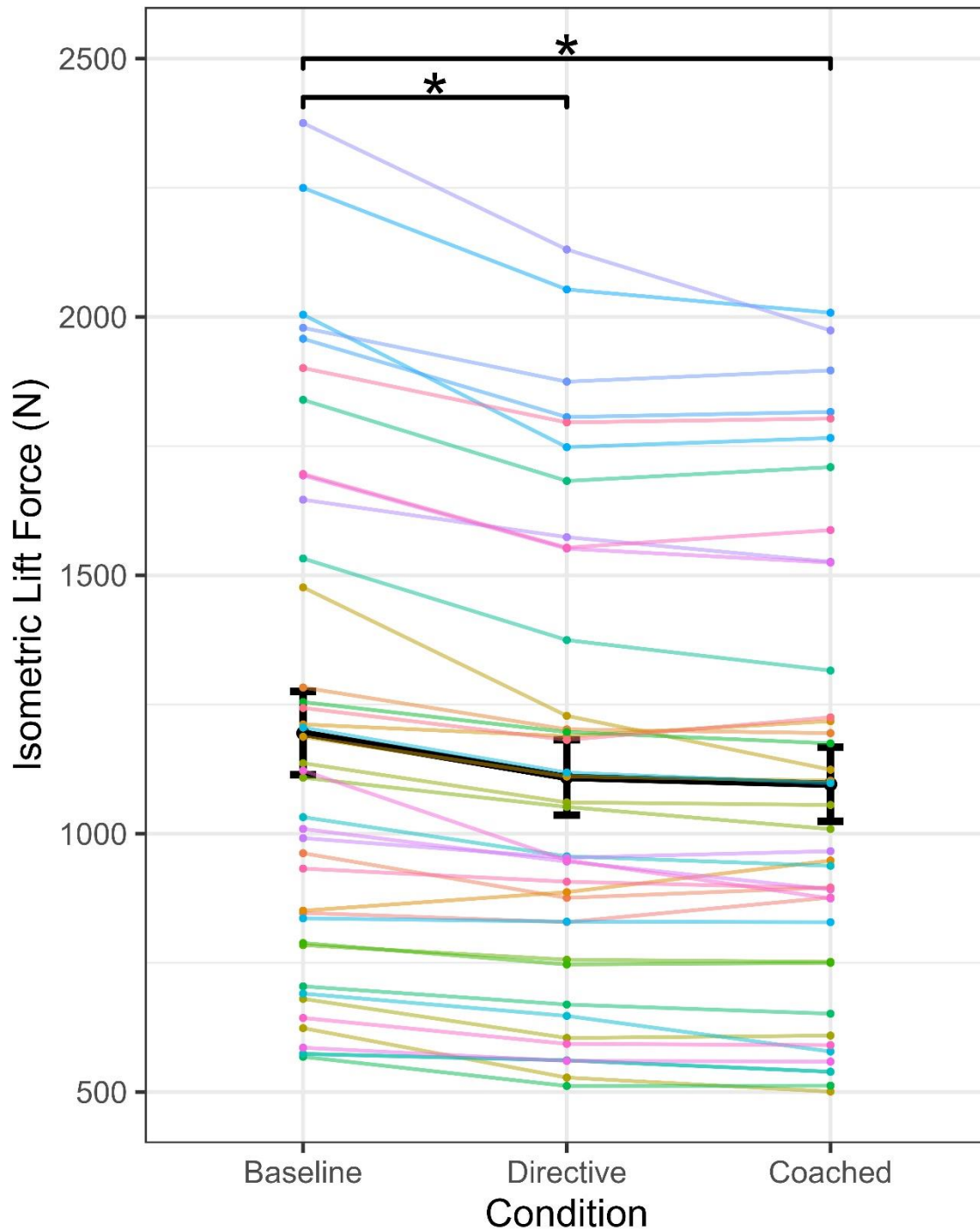


Figure 5.1: Force significantly decreased in the directive and coached condition compared to baseline but did not significantly differ between the directive and coached condition. Estimated marginal means are represented in black with the error bars representing standard error. The estimates for each participant from the linear mixed effects model are depicted by the coloured points and lines, where each colour represents a participant. Significant differences between conditions are indicated by “*”.

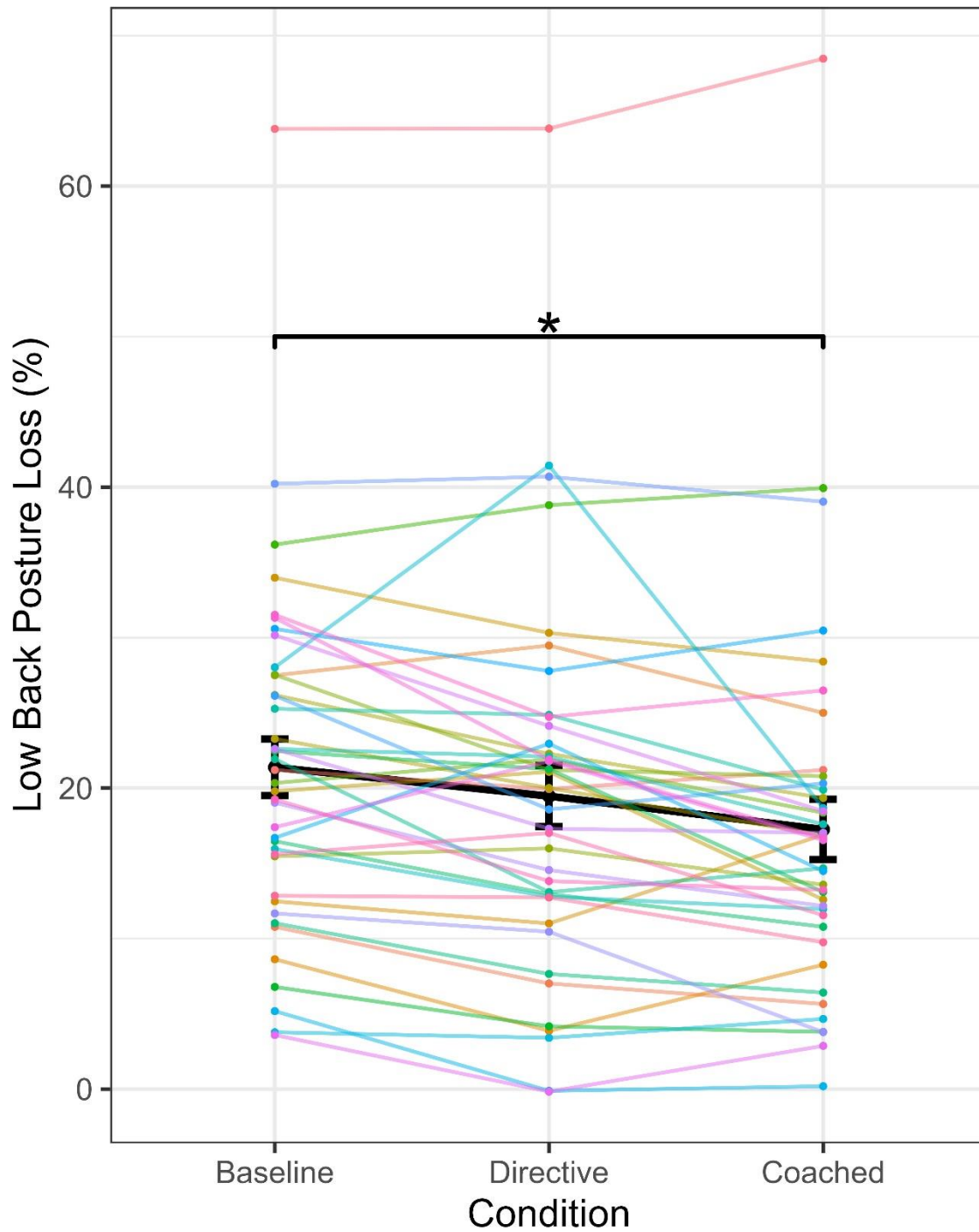


Figure 5.2: Low back postural displacement significantly decreased in the coached condition compared to baseline but did not significantly differ compared to the directive condition. Estimated marginal means are represented in black with the error bars representing standard error. The estimates for each participant from the linear mixed effects model are depicted by the coloured points and lines, where each colour represents a participant. Significant differences between conditions are indicated by “*”.

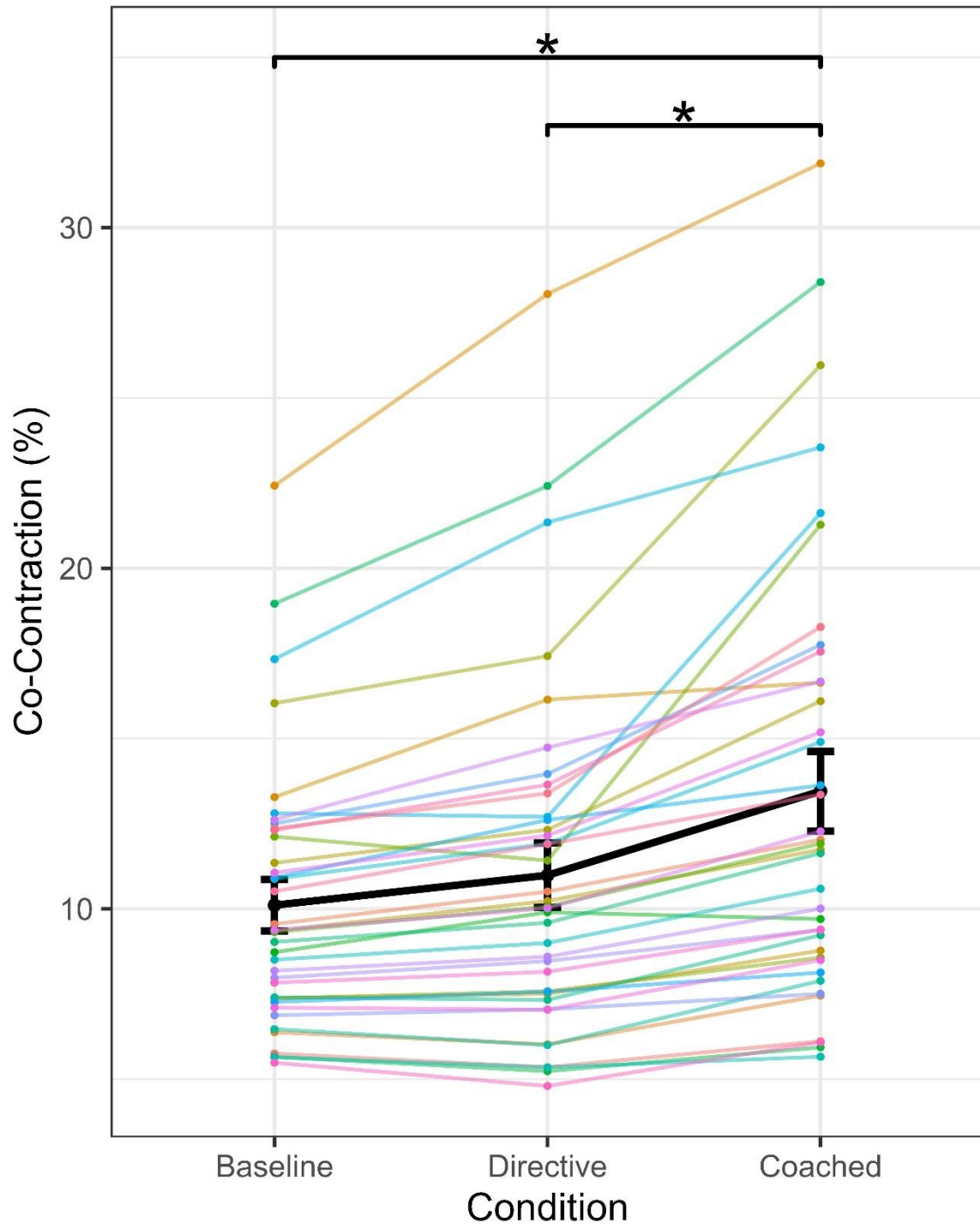


Figure 5.3: Trunk muscle co-contraction significantly increased in the coached condition compared to the baseline and directive condition, but the directive condition did not significantly differ from baseline. Estimated marginal means are represented in black with the error bars representing standard error. The estimates for each participant from the linear mixed effects model are depicted by the coloured points and lines, where each colour represents a participant. Significant differences between conditions are indicated by “*”.

Participants in this study varied in their physical activity history which ranged from those that do not resistance train to competitive powerlifters. When questioned if they typically thought about bracing or activating their core and how they cued themselves to do so, most participants reported that they typically braced or activated their core without consciously cueing themselves during activities. They described their strategy as “tightening” abdominal muscles or contracting their “core” (n=13), often with some form of breathing pattern which typically included inhalation (n=14). Other descriptions included inhalation without any description of muscle contraction (n=4) and contracting to draw the abdominals or umbilicus inwards towards the spine (n=6). Three participants reported that they do not think about or cue themselves on how they brace or activate their core. All participants commented that the coached technique was different from how they typically brace or activate their core. Most of the participants also preferred the coached bracing technique and reported that it made their trunk and/or body feel more “stiff”, “stable”, “solid” and/or “comfortable”. Some reported feeling stronger using the coached technique compared to the directive and baseline condition, while others commented that they felt like they could exert more force using the coached technique if given time and practice. Two participants did not prefer the coached technique as they felt it was too cognitively demanding. Other participants gave similar feedback, indicating that they had to give a high amount of attentional focus and cognitive effort to the detailed cues provided in the coached condition. Some noted they lost their brace and needed more practice using the coached condition to be able to maintain the brace while maximally exerting force.

5.4 Discussion

The coached detailed bracing technique was more effective than the simple verbal directive in increasing co-contraction and decreasing low back postural displacement compared to baseline. However, both types of cues decreased maximal isometric lifting force exertion. These results suggest that simple verbal directives are not as effective as detailed coaching at acutely modifying trunk muscle coordination to influence spine stiffness. Yet, cueing trunk muscle coordination during high force demand tasks can immediately inhibit force production regardless of the type of cue given.

The immediate decreases in peak lifting force after cueing trunk muscle coordination across both the directive and coached conditions may not be due to trunk muscle co-contraction. This is because isometric lifting force decreased to a similar extent in both the directive and coached conditions compared to baseline, but co-contraction only increased in the coached condition. Decreases in force may be better explained by the cognitive demand and attentional focus imposed by cueing trunk muscle coordination. Many participants reported that the coached condition required high amounts of effort to think about the detailed cues and attend to what they were doing. These results align with previous reports showing that cueing trunk muscle coordination can increase mental and physical workload (McPhee et al., 2017), and that cognitive demand can decrease force production (MacDonell & Keir, 2005; Mehta & Agnew, 2011). While the simple directive should not have required a cognitive effort as high as the coached condition, most participants reported that they typically activate their core automatically without needing to consciously focus on it. Thus, the instruction to “activate your core” may have created a cognitive demand to focus on the trunk muscles, which may have distracted from the

objective to produce maximal force. In both conditions, the resultant internal focus could have also interrupted the natural motor control to produce maximum force, as proposed by the constrained action hypothesis (Wulf, 2013). Thus, coaches choosing to cue trunk muscle coordination should do so with caution, when peak performance is the primary objective (e.g. during competition).

Some co-contraction is necessary for motor control (Gribble et al., 2003; McGill et al., 2003), but it is possible that co-contracting more than necessary may limit the contributions of the extensor moment to the net low back moment. If so, there should have been a difference in isometric force production between the directive and coached condition, as the directive condition did not elicit superfluous co-contraction for many individuals. Unfortunately, the current investigation may not clearly elucidate the effect of co-contraction on isometric end-effector force exertion due to the confounding influence of the cognitive effort, attentional focus and the difficulty participants reported in executing the coached bracing technique during the maximal exertion without additional practice. On average, the increase in co-contraction in the coached compared to the baseline condition was relatively low (Figure 5.3, Table 5.1), suggesting that cueing co-contraction may not substantially counter the net contributions of the extensor moments. However, this small change in co-contraction appears to be effective at resisting low back postural displacement

Although cueing trunk muscle coordination decreased maximal isometric force production, detailed coaching on how to brace but not simple verbal directives reduced low back postural displacement. This confirms that simple verbal directives are not as effective as the detailed coaching technique in reducing low back postural displacement.

The response to simple verbal directives can be attributed to prior beliefs concerning what it means to “activate your core”, which is evident from the various descriptions given by participants about what they typically do to brace or activate their core. Although the detailed coaching technique was expected to lead to more consistent results, the responses greatly varied across participants such that some increased low back postural displacement (Figure 5.2) and decreased co-contraction (Figure 5.3). This inter-individual variability in response can be explained by the fact that some participants found it difficult to immediately incorporate the coached technique during the maximal exertion and reported that they required more time and practice. Hence, using cues to modify trunk muscle coordination requires practice. Thus, investigations of the immediate responses to cueing different coordination strategies (hollow vs. brace) or using different types of cues (simple directive vs. detailed coaching) cannot be generalized until study designs capable of making inferences about acquisition, retention, and training transfer are used. Future studies should aim to provide more time for practice and adaptation so that the full extent of any potential effects can be observed without the confounding influence of learning, cognitive workload, and attentional focus that were observed in the current study.

A few other limitations of this work should be noted in addition to the limitations already discussed. This study is unique in that it provided and reported in detail the specific coaching cues used to cue participants to change their trunk muscle coordination. However, it did not quantitatively confirm if each participant learned the modified their trunk muscle coordination prior to observing its impact. In addition to providing more time and practice to learn any changes in trunk muscle coordination, future work should aim

to distinguish if participants are able to perform the brace as coached before evaluating any effects, as there can be high variability in participant responses as indicated in this study. Another limitation is that only one method of coaching trunk muscle bracing was investigated. It cannot be assumed that the approach taken in this study is the most appropriate way to coach trunk muscle coordination for all individuals, or for all tasks. Although the current results show that detailed coaching can immediately reduce low back postural displacement and increase co-contraction it remains unclear if increasing co-contraction can acutely enhance maximal force production due to the confounding effect of cognitive and attentional demands imposed by cueing trunk muscle coordination during the maximal exertion. The estimation of co-contraction assumed a constant EMG-to-force relationship across all the trunk muscle line of actions included in the musculoskeletal model and did not attempt to calibrate this relationship for each muscle which can impact the estimation of co-contraction (Cholewicki et al., 1995). The findings of the current study may have been influenced by fatigue as the data analyzed were from the end of a larger protocol, which included 19 maximal isometric lifting trials prior to the 9 trials analyzed here. Potential residual effects of coaching trunk muscle coordination demanded a sequential order of the experimental conditions which consequently makes it difficult to infer and account for the impact of fatigue due to the consistent order of conditions. Previous research has examined 56 (Kumar & Garand, 1992) and 90 (Chen et al., 2011) maximal isometric lifting exertions and did not report substantial influences of fatigue. Participants were asked about their fatigue and although some reported being fatigued compared to the start of the larger data collection, they did not feel like fatigue influenced them across the 9 trials analyzed here.

5.5 Conclusion

Detailed coaching is more effective than simple verbal directives at modifying trunk muscle coordination to increase co-contraction and decrease low back postural displacement. However, both detailed coaching and simple verbal directives can inhibit maximal isometric force production if provided immediately before a given exertion. To encompass the complete effects of modifying trunk muscle coordination, future work should go beyond investigating the immediate effects to provide time for learning the desired trunk muscle coordination patterns.

Chapter 6 : Explaining the Individually Varying Effects of Posture on Isometric Lifting Strength

6.1 Introduction

Posture can influence the ability to exert force in multi-joint tasks (Kumar & Garand, 1992), though its effect can vary considerably across individuals in magnitude and whether strength increases or decreases with a given postural change (Study 1, Chapter 4). For whole-body exertions like lifting, this suggests that a universally “strongest” postural profile may not exist, and that inferences about strength based solely on postures observed should be questioned. However, it is unclear if proxies or predictors of leverage that are derived from kinematic quantities, can reasonably capture the individual variation in the effect of posture on kinetic quantities like strength. If possible, it may then become feasible to make inferences about the effects of posture on strength in practice settings, as there are kinematic measurement techniques such as video recording that are simple to use and accessible. This could facilitate individualized movement assessment, coaching, and training in contexts when environment, equipment, and expertise to capture kinetic data in conjunction with kinematic data are limited.

The joint-to-external force horizontal distance (Figure 6.1 A and B) is one kinematic quantity that has been proposed as a proxy for external moment arm (Dai & Layer, 2019; Escamilla et al., 2000; Kipp & Harris, 2017; Pinto & Dickerson, 2021; Swinton et al., 2011, 2012; Van Den Tillaar & Ettema, 2009). Based on principles of leverage and mechanical advantage, this proxy has specifically been used to infer and conceptualize how muscular demand can vary with posture to influence force exerted at the end-effector in empirical

research comparing different postures (Dai & Layer, 2019; Escamilla et al., 2000; Kipp & Harris, 2017; Pinto & Dickerson, 2021; Swinton et al., 2011, 2012; Van Den Tillaar & Ettema, 2009) and in translational publications discussing the effects of joint posture on the ability to exert force (Keogh et al., 2013; Straub & Powers, 2024). Although, one study has shown that the glenohumeral joint-to-external hand force moment arm is associated with the maximal isometric hand force during upper extremity exertions in varying postures while seated (La Delfa & Potvin, 2016), it has yet to be verified if the joint-to-external force moment arms can explain variation in ability to exert force at the end-effector in more complex multi-joint tasks such as lifting a low-lying object.

In contrast to the joint-to-external force horizontal distance, the external moment arms effectively incorporated in formal analyses of mechanical linkages are calculated as the distances between the gravitational, inertial and inter-segment reaction forces acting on isolated body segments and an axis of rotation such as segment centre-of-mass or its end-points. If the assumptions of the joint-to-external force moment arm approach were adopted to ignore the effects of gravitational and inertial forces, the external moment arms of a joint relative to the inter-segment reaction force at the distal end-point of the segment (Figure 6.1 C) would conceptually represent a similar quantity as the joint-to-external force horizontal distance, but in the framework of a rigid linked segment model. Although this joint-to-inter-segment reaction force is not typically assessed, this representation of the external moment arm would be relative to the anatomical reference frame which would not be captured by the joint-to-external force moment arm. This potentially makes the joint-to-reaction force moment arm more sensitive to individually varying features such as body segment lengths, which have previously been associated with strength in multi-joint

tasks (Cholewa et al., 2019; Ferland et al., 2020; Lockie et al., 2018). However, quantifying the external moment arms alone, irrespective of the method discussed, may not sufficiently explain the effects of posture on the ability to exert force in all multi-joint tasks, as it ignores body segment mass-inertial properties and complex mechanical energy transfers within and between segments due to physical connections between adjacent segments and associated actions of uni- and multi-articular muscle-tendon complexes (van Ingen Schenau et al., 1994; Zajac et al., 2002). These factors are crucial in estimating the contribution of each joint to the end-effector force exerted which is evident from the differences between net joint moments estimated from rigid linked segment models which encompass some of these factors, and moments estimated from the product of the joint-to-external force moment arm and external force that ignore these factors (Toussaint et al., 1992; Wells, 1981).

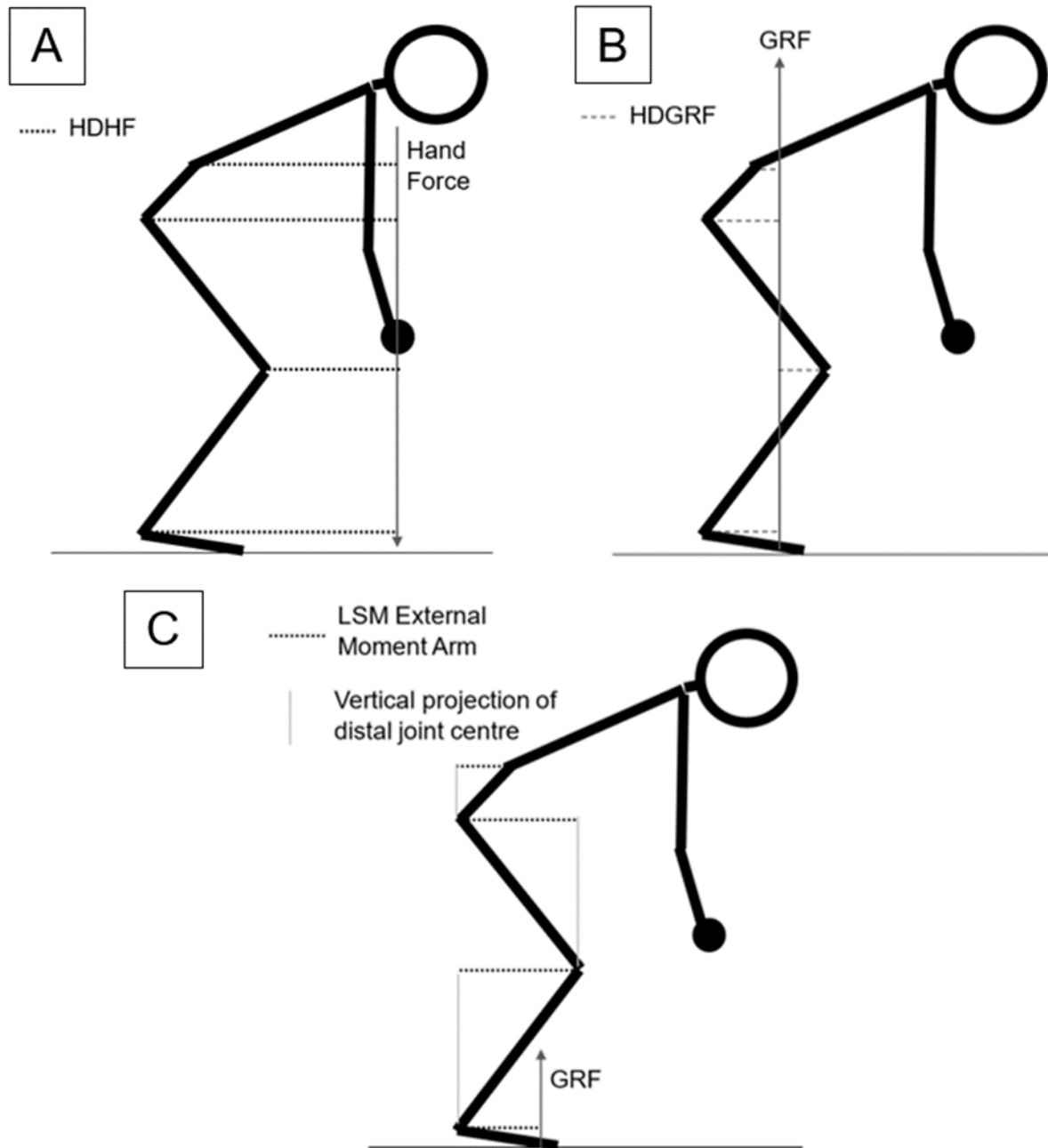


Figure 6.1: 2-dimensional depiction of the different external moment arm estimation approaches during a lifting task. A: Horizontal distance from the joint to the vertical projection of the hand force (HDHF). B: Horizontal distance of the joint to the vertical projection of the ground reaction force (HDGRF). C: Horizontal distance between the joint centre and the distal joint centre which marked the location of the inter-segment reaction force (HDRF).

A joint's torque-generating capacity is another proxy for leverage that has been predicted from kinematic quantities such as the joint angular position and instantaneous velocity. Joint angle modulates the muscle-to-joint moments arms (i.e. internal moment arms; Finni, 2006). Joint angle and velocity change the length and velocity of the musculotendinous unit which influences muscle force generation (Heckman & Sandercock, 1996; Herzog, 1996) and transfer across the musculotendinous unit (Finni, 2006; Heckman & Sandercock, 1996; Zajac, 1989). Yet, the effect of the interaction between the internal moment arms and musculotendinous unit length and velocity on joint torque are typically described as non-linear and can vary across directions of joint torques and across joints (Frey-Law et al., 2012; Loren et al., 1996; Zatsiorsky, 2003). As a result, normative data from strength testing has been used to model the joint angle- and velocity-dependent torque capacity of various joints to estimate the effects of movement execution on the ability to exert force (Anderson et al., 2007; Frey-Law et al., 2012; Hussain & Frey-Law, 2016; Stockdale, 2011). Digital human models have used the predicted joint torque-generating capacity to estimate the strength requirement for a given joint while performing a task (Abdel-Malek et al., 2019; Chaffin, 1997). Some researchers have simulated optimal postural profiles based on the predicted joint angle-velocity torque capacity, assuming that the postures and velocities providing highest joint torque capacities would maximize end-effector force exertion in multi-joint tasks (Xiang et al., 2019; Zaman et al., 2021). However, using predictions of joint torque-generating capacity based on normative data to estimate the effects of posture on the ability to exert force at the end-effector on an individual basis, may not capture potentially meaningful individual differences in the joint angle-velocity torque capacity (Alegre et al., 2014; Brughelli et al., 2010; Herzog et

al., 1991; Noorkõiv et al., 2014; Savelberg & Meijer, 2003) and in the relative strength between joints (e.g. knee relative to back strength; Li & Zhang, 2009). Hence, the extent to which normative data-based torque-generating capacity predictions can be used as a proxy for leverage on an individual basis is unclear.

The objective of this study was to examine the extent to which proxies for leverage derived from kinematic quantities can statistically explain the individual variation in the effects of posture on the ability to exert isometric force in multi-joint lifting. The joint external moment arms were calculated using 3 different approaches (Figure 6.1). The extent to which the external moment arms and predicted joint torque capacity could explain the variation in empirically measured maximal isometric lifting force across 4 different postural conditions was evaluated. The net joint reaction moments from a rigid link model were also evaluated to compare the extent to which variables derived from kinematic and kinetic data (e.g. net joint moments) could explain the individual varying effects of posture on isometric multi-joint strength. Consistent with fundamental biomechanical principles, it was hypothesized that the net joint reaction moments calculated using a rigid link model and inverse dynamics would explain a higher proportion of the individually varying effects of posture on the ability to exert isometric lifting force compared to the variables derived solely from kinematic quantities (external moment arms and predicted joint torque capacity).

6.2 Methods

This study analyzed data from Study 1 (Chapter 4), which motivated the current investigation.

6.2.1 Experimental Procedures

Participants performed 4 postural conditions in a block randomized order, which included 2 trunk inclinations (vertical and horizontal), each performed with 2 low back curvatures (neutral and flexed). For the trunk inclination conditions, participants were coached to adopt a lifting posture that enabled them to keep their trunk as vertical or horizontal as possible for each respective condition. Neutral low back curvature was cued to be within 30% of maximum low back flexion and the flexed curvature was cued to be above 80% of each individual's maximum low back flexion angle. Trunk inclination was modified through hip, knee, and ankle rotation, whereas low back curvature was modified by pelvis and ribcage rotation. Foot stance was kept consistent across the postural conditions. Real time optical motion tracking feedback and video were used to help participants learn and setup for each postural condition. Three trials were performed per condition with a minimum of 2 minutes of rest between trials (Study 1, Chapter 4).

6.2.2 Data Analysis

Since the postural conditions empirically observed primarily differed in posture of the lower limb and low back and the task was assumed to be symmetrical, this investigation focused on calculating each of the variables (external moment arms, predicted joint torque capacities, and net joint reaction moments) across the right ankle, right knee, right hip, and low back.

Kinematic and force plate data were dual pass low pass Butterworth filtered with a 10 Hz cutoff (Kristianslund et al., 2012; Tomescu et al., 2018), and used to calculate 3D joint centre positions, joint angles and net joint reaction moments for the right ankle, right knee, right hip and the low back in Visual 3D (Version 2022, HAS-Motion, Inc., Kingston, ON,

Canada). The joint centre of the low back was estimated as the proximal end of the pelvis segment at the mid-point between the iliac crests. A flexion/extension, lateral bend and axial twist Cardan angle rotation sequence was used to calculate joint angles relative to quiet standing. Net joint reaction moments were calculated using a linked segment model and bottom-up inverse dynamics using conventional Visual 3D anthropometric data scaled to each individual's height and body mass. Their data were exported from Visual 3D and imported into the subsequent analysis script in Python (Python Software Foundation, Wilmington, DE, USA; Spyder IDE, Version 5.3.0, Spyder Project Contributors).

The vector between the handle and load cell kinematic marker was interpolated to 2000 Hz and then used to find the vertical component of the lifting force recorded by the load cell. The peak vertical lifting force and the instant it occurred was identified. The 3D joint centre locations, joint angles and net joint reaction moments of the right ankle, right knee, right hip, and low back at the instant of peak vertical lifting force were identified. The external moment arms were calculated using 2 variations of the joint-to-external force moment arm approach and 1 variation of the joint-to-inter-segment reaction force moment arm (Figure 6.1):

- 1) Horizontal distance to the hand force (HDHF): Previous research has estimated the joint-to-external force moment arm as the horizontal distance between a joint a barbell held by the hands as the location of the external force (Escamilla et al., 2000; Pinto & Dickerson, 2021; Swinton et al., 2011, 2012; Van Den Tillaar & Ettema, 2009). Thus, the first variation of the joint-to-external force moment arm investigated was the horizontal distance between the joint and the vertical

component of the external end-effector force exerted at the hands. The location of the hand force was marked by the kinematic infrared emitting diode placed on the middle of the handle (Figure 6.1 A).

- 2) Horizontal distance to the ground reaction force (HDGRF): The second variation of the joint-to-external force moment arm within the published literature is the perpendicular distance between the joint and the projection of the ground reaction force (Kipp & Harris, 2017). The location of the ground reaction force was taken as the centre of pressure of the right foot (Figure 6.1 B).

Since the task investigated was maximal isometric lifting exertions, the vertical component of the exerted force was of primary interest. Calculating the horizontal distance between the joint and the location of the external force is equivalent to estimating the perpendicular distance between the joint centre and the vertical component of the external force vector. Thus, the horizontal distance between each joint and each of the 2 external force locations (HDHF and HDGRF) were calculated using the anterior posterior and mediolateral axes coordinates:

$$\text{Joint - to - External Force Horizontal Distance} = \sqrt{(x_{\text{joint centre}} - x_{\text{External Force}})^2 + (z_{\text{joint centre}} - z_{\text{External Force}})^2}$$

Where x and z are respectively the anterior posterior and mediolateral axes coordinates for the given joint and for the location of the external force.

- 3) Horizontal distance to the inter-segment reaction force (HDRF): The joint-to-inter-segment reaction force moment arm was estimated as the horizontal distance between the joint centre and the location of the inter-segment reaction force which represents the effect of the external force on a given segment was calculated as:

*Joint – to – Reaction Force
Moment Arm*

$$= \sqrt{(x_{\text{joint centre}} - x_{\text{Reaction Force}})^2 + (z_{\text{joint centre}} - z_{\text{Reaction Force}})^2}$$

Where the location of the vertical component of the inter-segment reaction force for the ankle joint was the centre of pressure of the force plate, and the location of the reaction force for all other body segments was the joint centre of the distal segment.

The joint angle dependent torque capacity was estimated using the logistic equation proposed by Looft Frey-Law (2015):

$$S = \frac{4 C1 e^{\frac{-(\theta-C2)}{C3}}}{(1 + e^{\frac{-(\theta-C2)}{C3}})^2} + \frac{4 C4 e^{\frac{-(\omega-C5)}{C6}}}{(1 + e^{\frac{-(\omega-C5)}{C6}})^2} + \frac{16 C7 e^{\frac{-(\theta-C2)}{C3} - \frac{(\omega-C5)}{C6}}}{(1 + e^{\frac{-(\theta-C2)}{C3}})^2 \times (1 + e^{\frac{-(\omega-C5)}{C6}})^2}$$

Where S is the estimated maximum torque generated by a joint, as a function of joint angle (θ) and joint angular velocity (ω). C1-7 are the coefficients specific to each

joint (ankle, knee, hip, trunk), direction (flexion or extension) and sex, which were determined by fitting the logistic equation to previously reported mean maximal torque data from strength testing using a Levenberg-Marquardt algorithm least squares optimization (Frey-Law et al., 2012; Hussain & Frey-Law, 2016; Looft & Frey-Law, 2015; Stockdale, 2011). Since modifying posture has been shown to change the direction of the net joint reaction moments (Hwang et al., 2009; Toussaint et al., 1992), the calculated net joint reaction moment at the instant of peak vertical lifting force for each joint was used to determine the direction of torque generation. Specifically, the direction (flexion/extension) was selected to oppose the net joint reaction moment such that a flexor net joint reaction moment would require joint extension torque. Only the knee net joint moment changed in direction within the current data across the different postures. Accordingly, the coefficients for flexion or extension maximum knee torque were used depending on the direction of the net joint reaction moment. The velocity was assumed to be 0°/s as the task was isometric.

6.2.3 Statistical Analysis

Linear mixed effects models were used to analyse the relationship between the observed peak vertical force exerted on the handle (dependent/outcome variable) with each biomechanical variable calculated (independent variable/fixed effect; lmerTest package, RStudio Version RStudio 2023.03.0, The R Foundation, Vienna, Austria). The biomechanical variables were 3 different calculation variations of the external moment arm, the predicted joint torque capacity, and the net joint reaction moments. A linear mixed effects model was created for each biomechanical variable and the respective

variable values for the right ankle, right knee, right hip, and low back were each entered as individual independent variables within each model.

The linear mixed effects model allowed for the regression analysis to account for individual regression slopes for each participant by fitting the model with a random intercept for each participant and a random slope for the relationship to each biomechanical variable across the observed joints. The marginal and conditional R^2 values were estimated to determine the extent to which each biomechanical variable could explain the variation in the effects of posture on observed maximum vertical lifting force across all individuals (Nakagawa et al., 2017). The marginal R^2 represents the proportion of total model variance explained by the fixed effects which are the aggregate regression estimates (i.e. group level estimate). The conditional R^2 represents the proportion of variance explained by both the fixed effects and the random effects which represent the between individual differences in the fixed effect estimates. Accordingly, the marginal R^2 indicates the extent to which the estimated fixed effect (i.e. aggregate group level) relationship between the values calculated for the respective biomechanical variable across the right ankle, right knee, right hip, and low back, could explain the total variance in the model which includes the individually varying effects of posture on isometric lifting strength. The conditional R^2 value indicates the extent to which considering the random effects (i.e. varying intercepts and slopes for each participant) together with the fixed effects would explain the total variation in the model. Thus, a high marginal R^2 value would indicate that the biomechanical variable would capture the individually varying effects of posture on isometric lifting strength.

6.3 Results

The fixed effect (aggregate) relationship between the biomechanical variables derived from kinematic measurements (external moment arms and predicted joint torque capacities) explained a very low proportion of the variance in the effects of posture on the maximal isometric lifting force, as indicated by the marginal R^2 values (below 17%; Table 6.1). Accounting for the heterogeneity in the relationship between the external moment arms or the predicted joint torque capacities with the exerted isometric lifting force, by estimating intercepts and slopes for each participant, increased the ability of these biomechanical variables to explain the total variance in the model, as indicated by the higher conditional R^2 values (greater than 93%; Table 6.1). Thus, the external moment arms and predicted joint torque capacities are sensitive to the effect of posture on the ability to exert maximal isometric force, but do not consistently capture individual differences in the effect of posture on isometric lifting strength.

In contrast, the fixed effect relationship between the net joint reaction moments and isometric lifting strength was able to explain a high proportion of the variation in the model as indicated by the marginal R^2 values (approximately 80%; Table 6.1). This suggests that the individually varying effects of posture on the ability to exert maximal isometric force in a multi-joint task are more consistently explained by the net joint reaction moment estimated from formal rigid linked segment modelling, without needing to estimate individual differences in the relationship between the net joint reaction moment and the exerted force.

Table 6.1: Marginal and conditional R^2 values calculated from linear mixed effects models for each biomechanical variable investigated. The marginal and condition R^2 describe the proportion of variance explained by only the fixed effects (aggregate/group level relationship) and both the fixed and random effects (individually varying relationships) respectively.

Biomechanical Variable	Marginal R^2	Conditional R^2
HDHF	0.123	0.958
HDGRF	0.018	0.941
HDRF	0.035	0.944
Torque Capacity	0.169	0.930
Net Joint Moment	0.795	0.986

Note: HDHF: horizontal distance to the hand force, HDGRF: horizontal distance to the ground reaction force. HDRF: horizontal distance to the inter-segment reaction force.

6.4 Discussion

The external moment arms and predicted torque capacity which are derived from kinematic quantities, were not able to explain the variation in the effects of posture on isometric lifting strength. Although these variables may encompass relevant information, they did not independently capture the varying effects of posture on isometric lifting strength without accounting for individual differences in the relationship between these variables and isometric lifting strength. In contrast and as expected based on fundamental biomechanical principles, the net joint reaction moment explained a high proportion of the total variation in the effects of posture on isometric lifting strength without needing to independently account for any individual differences in the effect of the net joint reaction moments on isometric lifting strength. Thus, variables derived solely from kinematic data may not be the best suited for assessing the individually varying effects of posture on isometric force exerted at the end-effector in a multi-joint task.

The current results suggest that the external moment arms alone cannot capture the effects of posture on the ability to exert force in multi-joint tasks. Thus, inferences based solely on the changes in external moments arms due to postural modifications should be questioned. Additionally, it appears that the joint-to-inter-segment-reaction force does not provide any substantial improvement over the joint-to-external force in capturing the effects of posture on the ability to exert force. The inability of the external moment arms to serve as a proxy for leverage in multi-joint tasks can be expected based on fundamental biomechanical principles. Net joint moments are influenced by more than just external moment arms, as there are several forces and moments that also contribute. For example, in Figure 6.2 the equation to calculate the net joint moment about the z axis of a generic segment relative to its centre of mass is:

$$\begin{aligned}\sum M_z &= I_z \alpha_z \\ &= r_{prox\ x} R_{prox\ x} + r_{prox\ y} R_{prox\ y} + r_{dist\ x} R_{dist\ x} + r_{dist\ y} R_{dist\ y} \\ &\quad + M_{prox\ z} + M_{dist\ z}\end{aligned}$$

Where M_z is the moment about the z axis, I_z is the moment of inertia about the z axis, α_z is the angular acceleration in the z axis, r is the joint-to-reaction force moment arm with respect to the segment centre of mass, R_x and R_y are the reaction forces in the x and y axes, and subscripts *prox* and *dist* refer to the proximal and distal end-points of the segment. Figure 6.2 and the above equation highlights the impact of features such as

mass-inertial properties and the between segment coupling of forces and moments to highlight the limitations to making inferences solely on the external moment arms. Theoretically, the resultant forces exerted at the end-effector should be the sum of all the net joint moments across the linkage and due to the multiple factors involved, these resultant forces at the end-effector cannot be deduced from the external moment arms alone. Apart from one study showing that the joint-to-external force moment arm of the shoulder was correlated to the hand force in seated upper arm maximal isometric exertion tasks (La Delfa & Potvin, 2016), no known study has evidenced a relationship between the external moment arms and end-effector force production in multi-joint tasks. It is possible that the external moment arm may indicate the effects of posture on force in certain simplified contexts such as this seated upper extremity task, where the effects of mass-inertial properties may be trivial. However, in more complex multi-joint tasks such as lifting an object from the ground, the mass-inertial properties of the body and the inter-segment reaction kinetics cannot be ignored as evident from the current results and discussion.

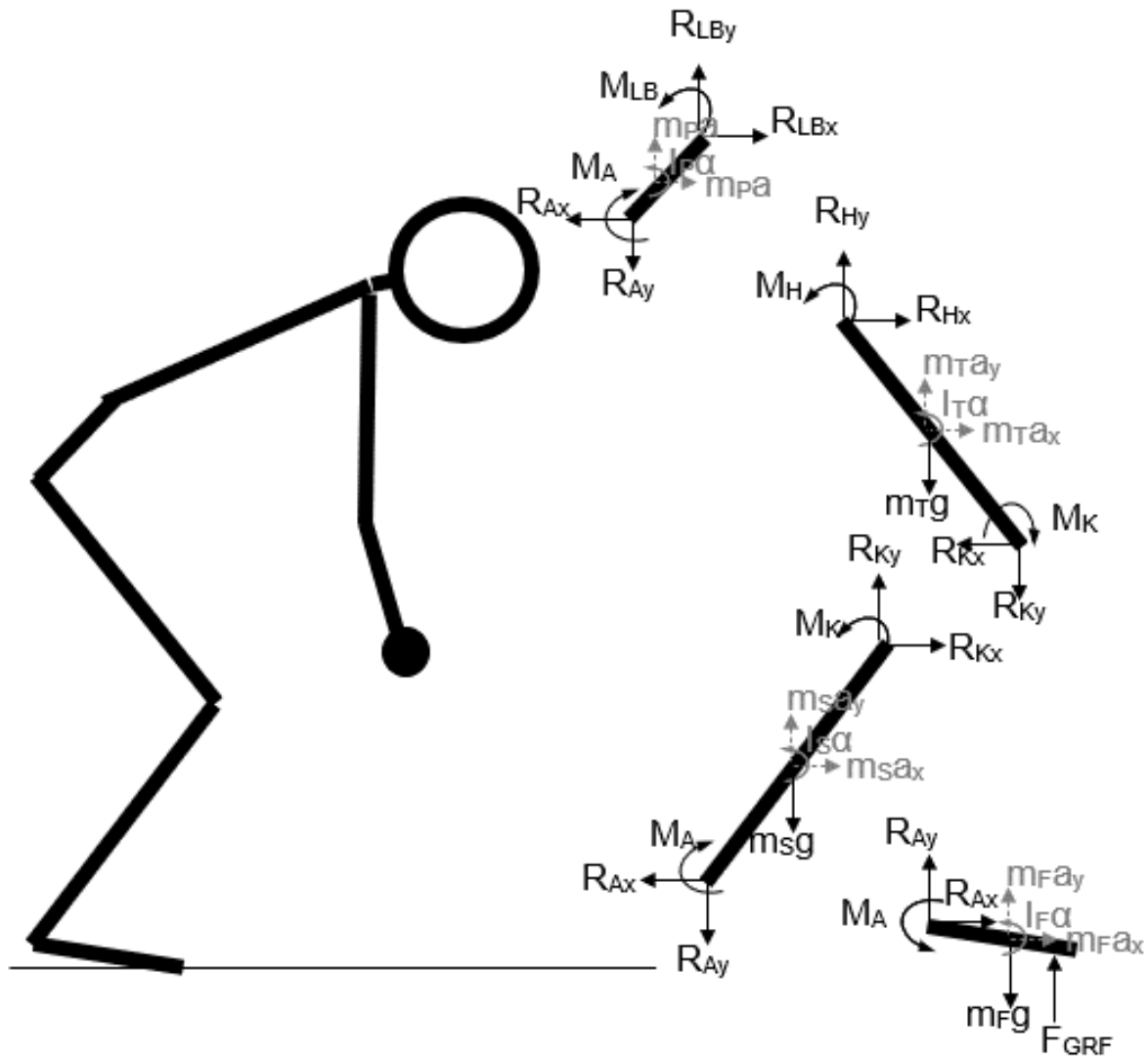


Figure 6.2: Free body diagrams for the foot, shank, thigh, and pelvis segments (subscripts F, S, T, and P respectively), depicting the ground reaction force (FGRF) as well as the inter-segment reaction forces (R_x and R_y) and net joint reaction moments (M) at the ankle, knee, hip, and low back (subscripts A, K, H, and LB respectively).

Similarly, the predicted joint torque capacities were not able to explain the individually varying effects of posture on isometric force exertion. Several nuances should be considered with respect to this. One consideration is whether the joint angle torque

relationship can be generalized across individuals. Differences between sex were considered in the joint-angle-torque relationship by using sex specific model coefficients to predict joint torque capacity. However, individual differences that cannot be explained by sex may need to be considered as individuals can vary in the relative strength between joints (e.g. knee relative to back strength; Li & Zhang, 2009), and in the joint angle-torque relationship due to adaptations from previous training or activities (Alegre et al., 2014; Brughelli et al., 2010; Herzog et al., 1991; Noorkõiv et al., 2014; Savelberg & Meijer, 2003). Thus, improving the prediction of joint torque capacity to encompass individual differences may advance the ability to assess the effects of posture on the ability to exert force on an individual basis. Additionally, torque-generating capacity was predicted from single joint dynamometer testing data which facilitates control of posture and velocity when modelling the joint angle-velocity-torque relationship. This prediction based on single joint strength testing datasets also do not account for adjacent joint interactions such as how the joint angle-torque relationship can vary with postural (Kellis & Blazeovich, 2022) or muscular coordination (McGill, McDermott, et al., 2009) changes at adjacent joints. An alternative method to model joint posture dependent strength has been through modelling the relationship between posture, anthropometrics, and the peak net joint moments during isometric multi-joint strength test such as lifting, pulling and pushing (Chaffin, 1997). This alternative approach has shown to be effective at predicting population strength values (i.e. mean group strength), but it remains uncertain if it is effective at capturing the effects of posture on strength on an individual basis.

In contrast to the variables derived solely from kinematic measurements, the net joint moments calculated using inverse dynamics and a 3D rigid linked segment model

was able to explain a high proportion (approximately 80%) of the variation in the effects of posture on isometric lifting strength. Although this does not solely encompass individually varying effects of posture on the ability to exert force, it suggests promise for using variables derived from both kinematic and kinetic variables as this facilitates an analysis that encompasses body segment mass-inertial properties and the complex mechanical energy transfers within and between segments. The net joint moment of multiple joints can also be relatively easier to consolidate into a single variable or score compared to the external moment arms, as the sum of the net joint reaction moment across multiple joints has shown to be associated with external work and power performed by the body (Flanagan & Salem, 2005). Although the magnitude of the net joint reaction moment may covary with the end-effector force exerted, the distribution can differ within an individual, across task conditions (Flanagan & Salem, 2008). As opposed to using variables solely derived from kinematic data, it may be of benefit to further develop techniques to use the net joint moment to assess the individually varying effects of posture on isometric strength to create approaches that can assess the effect of movement on physical performance on an individual basis. Doing so can enhance the ability to modify movement to preserve musculoskeletal health while enhancing physical performance as a single movement profile cannot be generalized to maximize each of these objectives across all individuals due to the variation in the effects of posture on the ability to maximally exert force (Study 1, Chapter 4). It is possible that other biomechanical or non-biomechanical factors may also need to be considered. For instance, the balance control demands have previously been suggested to also contribute to influencing the potential to exert force (Fischer et al., 2013; Kerk et al., 1994). Further investigation will reveal the

extent to which an individual's movement profile can be indicative of their ability to exert force.

This work is limited in that it did not thoroughly identify how the biomechanical variables investigated can best be used to assess the individually varying effects of posture on the ability to exert force. It only assessed whether the variables investigated encompassed relevant information to explain the individually varying effects of posture on the ability to exert force. Since the experimental postural conditions primarily modified lower limb and low back posture and were assumed to do so symmetrically, the biomechanical variables investigated were only quantified for the right ankle, right knee, right hip, and low back. Although this assumption limited the analysis from capturing all the effects of postural changes across the body that may have been of consequence to the exerted lifting force, the assumption made may be appropriate as the net joint reaction moment of the assessed joints were able to explain a high proportion of the variance in the effects of posture on isometric lifting strength. Given the scarcity of research, the current work informs the development of strategies to assess how movement can influence the ability to exert force in multi-joint tasks. Since postural conditions were experimentally imposed, it is difficult to expect that every participant would be equally proficient to produce maximal force in each condition, on every trial, while maintaining the target posture (Farrow & Robertson, 2017; Pinto & Dickerson, 2021). Using isometric as opposed to dynamic exertions were selected to facilitate the maintenance of targeted postures while exerting maximal effort. However, some participants lost joint posture as they exerted maximal force (Study 1, Chapter 4). Not every trial collected (3 trials per condition) may represent the maximal force exertion capability for each postural condition

due to potential learning and adaptation that may have occurred within each experimentally imposed postural condition and may contribute to some of the unexplained variation in the statistical models investigated. Additional practice and trials may have facilitated a better representation of posture's effect on the ability to maximally exert force. Yet, this would be at the cost of fatigue. This supports a need to investigate the consistency of multiple maximal isometric lifting trials and the extent that learning and fatigue could have influenced these measurements.

6.5 Conclusion

Joint external moment arms and predicted joint torque capacities did not explain the individually varying effects of posture on the ability to exert force. In contrast, variables derived from both kinematic and kinetic measurements such as the net joint reaction moments, calculated using a rigid linked segment model, appear to demonstrate promise in capturing the individually varying effects of posture on isometric lifting strength. Future work should determine how to use the net joint reaction moment to assess the effects of movement on physical performance and subsequently modify movement to enhance physical performance.

Chapter 7 : Impact of Individual Responses to Postural Directives on Isometric Lifting Strength

7.1 Introduction

Although reducing the demands of an occupational task (e.g. weight lifted) can reduce the forces imposed on the musculoskeletal system (Gooyers et al., 2018), personal patterns of movement coordination and control can counteract modifications of task demand (Faber et al., 2007; McGill, 2009). Specifically, movement behaviour regulates the relationship between external task demands and applied tissue loads while also influencing tissue load tolerance and pain perception (Cholewicki et al., 2019; Marras et al., 2016; McGill, 2009; Nelson & Hughes, 2009). For example, low back flexion migrates the distribution of internal forces from active to passive tissues (Potvin, McGill, et al., 1991; Zwambag & Brown, 2020), which have reduced capacity and require longer durations for recovery and adaptation (Buckwalter & Grodzinsky, 1999; Kannus et al., 2007; Vila Pouca et al., 2021). Low back flexion reduces the tolerance of spine tissues to loading (Callaghan & McGill, 2001; Gallagher et al., 2005; Gunning et al., 2001; Howarth & Callaghan, 2012; Solomonow, 2012) while diminishing the ability of the back extensor muscles to counteract potentially injurious anterior shear forces (McGill et al., 2000; Pinto et al., 2021). Repetitive and sustained low back flexion also imposes residual effects through viscoelastic changes that disrupt sensorimotor function (Abboud et al., 2018; Pinto et al., 2021; Sánchez-Zuriaga et al., 2010; Solomonow, 2012) and pain perception (Viggiani & Callaghan, 2021, 2022), which can potentially contribute to LBD. Thus,

modifying movement behaviour by targeting features such as low back flexion may enhance efforts to reduce biomechanical loading and LBD.

It is vital that modifying movement behaviour to manage biomechanical loading does not negatively impact physical performance, as this may limit acceptability, sustainability, and effectiveness of the modifications. For instance, manual materials handling training has previously targeted lifting with a squat technique to maintain a vertical trunk inclination (Denis et al., 2020; van Dieën et al., 1999). Apart from the questionable impact on reducing biomechanical loading (van Dieën et al., 1999), adopting a squat lifting technique can decrease strength (Kumar & Garand, 1992), increase energy expenditure (Garg & Herrin, 1979; Kumar, 1984; Welbergen et al., 1991), be perceived to be more exerting (Straker & Duncan, 2000) and reduce the ability to balance (Chow et al., 2005; Heiss et al., 2002). These effects of posture on physical performance can hinder task performance and have been proposed as plausible explanations for the poor adherence to and ineffectiveness of lifting technique training (Denis et al., 2020). Thus, strategies to assess and modify movement to manage biomechanical loading should also consider the concurrent effects on physical performance.

As opposed to targeting a general lifting technique that imposes a single vaguely defined whole-body postural profile that might not be suitable for every context or situation, targeting a specific movement feature such as low back posture in multi-joint tasks may offer more opportunity for self-organization and movement optimization. That is, performers would be afforded greater flexibility to configure their body linkage in ways that enable them to satisfy task objectives if fewer instructional constraints are imposed. Prior research employing this “key feature” approach has demonstrated promise as a way

to modify low back biomechanical tissue loading and tolerance while affording opportunity to self-organize the rest of the linkage (Beach et al., 2018; Becker et al., 2022; Chan et al., 2022; Frost et al., 2015; Grütters et al., 2023; Larson & Brown, 2023; Pinto et al., 2018). However, there is evidence that limiting low back flexion during maximal effort isometric lifting exertions acutely reduces ability to generate end-effector force in some individuals (Study 1, Chapter 4), which may initiate and accelerate fatigue-related processes in those affected. Within the context of low demand tasks, cueing specific movement features using simple directives, without explicitly coaching individuals to accommodate immediate effects of cueing on physical performance potential, should thus be questioned. If the task demands do not necessitate prioritizing acute physical performance, it cannot be assumed that individuals will self-organize in a manner that maintains their potential to physically perform. To our knowledge, no study has explicitly assessed how physical performance is impacted by cueing low back posture during sub-maximal tasks such as those typically performed during work or daily activities.

The objective of this study was to observe how an instruction to avoid rounding the low back during light mass lifting immediately impacts isometric lifting strength of the adopted posture. Participants lifted a crate without (baseline) and with the instruction to avoid rounding the low back. Posture at the initiation of a crate lift for each condition (baseline and instruction) was replicated to perform maximal isometric lifting exertions to determine how postural responses impacted strength. Given results of Study 1, it was hypothesized that not every individual would respond to the simple verbal instruction during light mass lifting in a manner that maintains strength.

7.2 Methods

7.2.1 Experimental Procedures

A schematic overview of the experimental procedures for this study is provided in Figure 7.1. Maximal isometric lifting exertions were performed to estimate lifting strength. A minimum of 3 warm-up isometric lifting trials were performed where participants were instructed to exert 30, 50 and 80 % of what they perceived to be their maximal effort. This was followed by 3 maximal isometric lifting trials in a self-selected posture. The unit vector between the kinematic markers on the handle and force transducer was multiplied by the force recorded by the force transducer to find the vertical component of the exerted lifting force. A crate was loaded with weight approximating 10% of the maximum vertical force exerted across these 3 trials in the self-selected body posture, for subsequent trials.

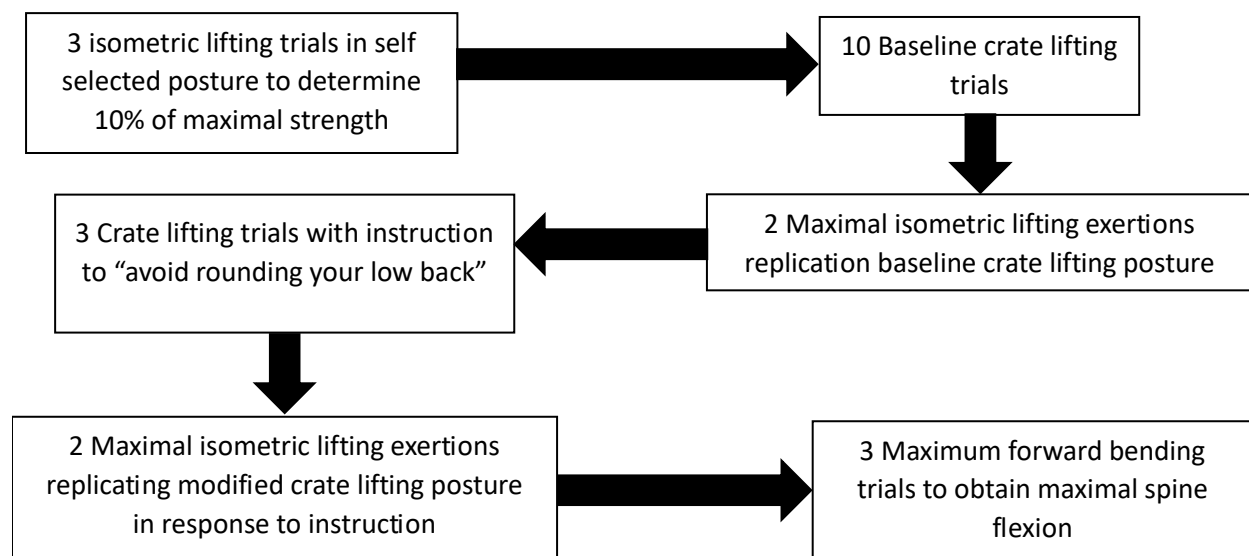


Figure 7.1: Experimental procedure flowchart.

The loaded crate was placed on a platform so it could lay flat over the metal bar on the ground where the lifting handle was secured (Figure 7.2). The crate was lifted for 10 separate baseline trials where the only instruction was “to lift naturally and as you regularly would in everyday life”. For each crate lifting trial, participants started from standing, reached and lifted the crate from the platform to a standing position, lowered the crate, and finally returned to the standing position without the crate. The crate was positioned in the same location on the platform before every trial to facilitate consistency in experimental environment and replication of lifting posture for the maximal isometric lifting trials.

Two maximal isometric lifting exertion trials were performed in a posture replicating the instant that the participant initiated lifting the crate from the platform. To accomplish this, a picture of the sagittal plane of the participant at the approximate instant they lifted the crate up from the platform was captured on the 10th baseline crate lifting trial (Figure 7.2A). Participants were instructed to maintain their foot position until the 2 isometric trials were completed. The attachment point of the isometric lifting handle on the metal bar that secured it to the ground, was moved to align with the centre of the base of the crate on the platform. The picture of the participant initiating the lift was used to draw a silhouette of the participant’s whole-body posture on a transparent sheet which overlay a computer screen. The computer screen was in view of the participant and displayed real-time video of the participant so they could match their posture to the silhouette. The range in peak joint flexion/extension angles of the low back, hips and knees from the 8th, 9th, and 10th baseline trials together with real-time outputs of these joint angles (NDI First Principles software Version 1.2.4, Northern Digital Inc., Waterloo, ON, Canada), were used to verify

that participants were within range of their posture at the initiation of lifting the crate. Once the posture was replicated and participants confirmed they were ready, a 3 second countdown began after which participants exerted maximal isometric lifting effort. Prior to these exertions, instruction was given to maintain setup posture during the maximal exertion.

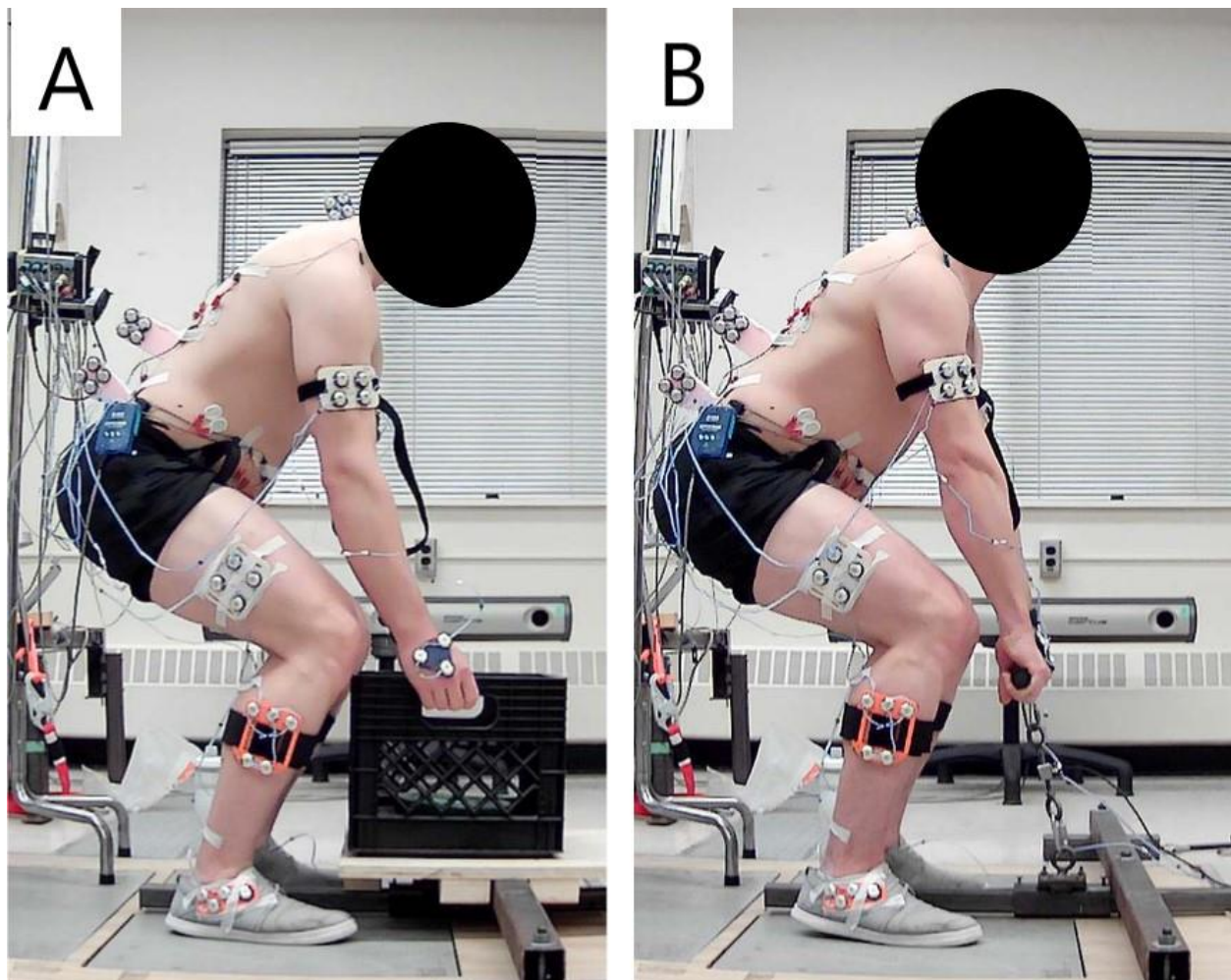


Figure 7.2: (A) Example of picture taken at the initiation of crate lift which was used to replicate posture. (B) Participant exerting maximal effort during isometric lifting trial in the replicated posture.

Three crate lifting trials were then performed with only the instruction to “avoid rounding your lower back”. The researcher (BLP) also briefly visually demonstrated the difference between flexion and extension of the low back while indicating that “rounding” referred to a flexed curvature. No other explanation, description, or coaching was given regarding low back posture or how to lift. This was followed by 2 maximal isometric lifting trials replicating the posture at the instant that the participant initiated lifting the crate from the platform during the last lifting trial of the instruction condition. The procedures for replicating this posture were identical to those used for the baseline trials.

Finally, 3 trials of standing maximum forward bending were performed. For each trial, participants were coached on rotating their pelvis and ribcage to achieve maximum low back flexion as they bent their chest towards their thighs. At volitional end range of low back flexion, a 3 second trial was collected during which participants were encouraged to flex their low back further. These trials were performed after the experimental conditions to avoid interfering with participants’ natural response and ability to modify low back posture.

A minimum of 2 minutes of rest were taken between maximal isometric lifting trials.

7.2.2 Data Analysis

Kinematic data were processed in Visual3D (Version 2022, HAS-Motion, Inc., Kingston, ON, Canada) and filtered with a dual pass Butterworth low pass filter with a 10 Hz cutoff (Kristianslund et al., 2012; Tomescu et al., 2018). Low back, hip, knee and ankle 3D joint angles were calculated relative to relaxed standing, using a flexion/extension, lateral bend and axial twist Cardan angle rotation sequence. These data were imported in the

subsequent analysis performed in Python (Python Software Foundation, Wilmington, DE, USA; Spyder IDE, Version 5.3.0, Spyder Project Contributors).

The analysis of the crate lifts began with isolating the time period of the lifting phase by identifying the start and end of the lifting phase (Figure 7.3). The start was marked by the onset of bending movement from a standing position towards the crate and the end was marked as the onset of the bending movement from standing to lower the crate. These onsets were identified using the first derivative approach and the right hip flexion/extension joint angle (Pinto & Callaghan, 2023). Specifically, the start was defined as the first instant of the longest time period where the first derivative was greater than 5 times the standard deviation of 1.5 seconds of relaxed standing captured at the start of each trial. This time instant marked a continuous increase in hip flexion as the participant bent down to lift the crate. The detection of the lowering phase onset began from the time point after the start of the lifting phase where the hip extended to be within 60% of the maximum hip extension during the trial. The onset of the lowering phase (i.e. end of the lifting phase) was marked as the first instant after this time point where the first derivative was positive for the longest duration up to the next peak in hip flexion. The maximum low back flexion, right hip flexion and right knee flexion achieved during the identified lifting phase of each trial was extracted and normalized to participants' maximum low back flexion range-of-motion.

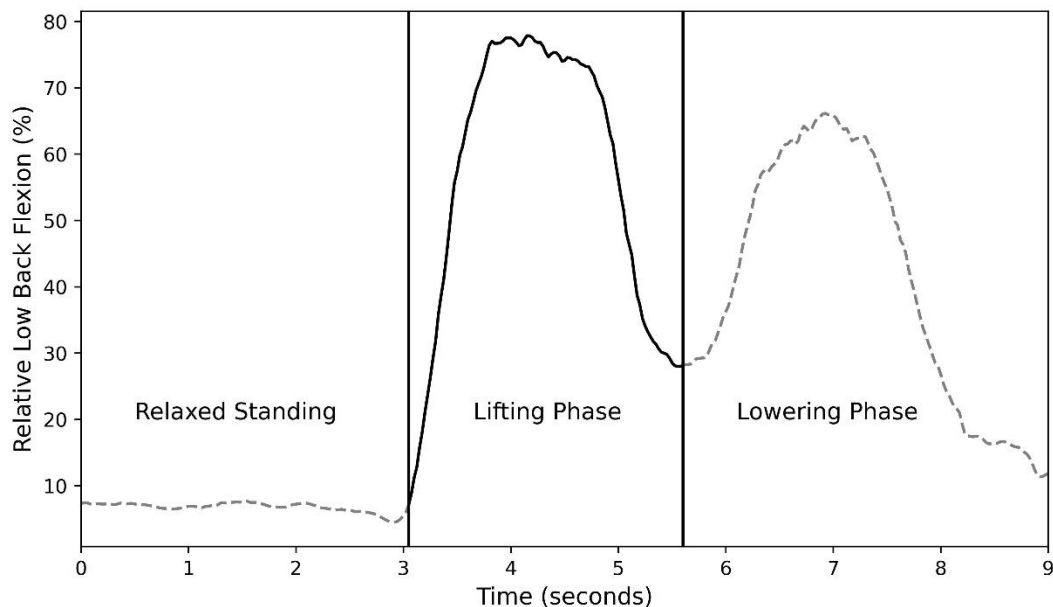


Figure 7.3: Sample participant data depicting the identified lifting phase plotted using a solid black line. The vertical solid black lines represent the start and end of the lifting phase.

The analysis of the maximal isometric lifting exertions involved interpolating the unit vector between the handle and force transducer to 2000 Hz to find the peak vertical component of the force exerted on the handle.

7.2.3 Statistical Analysis

Linear mixed effects models were used for group statistical analyses for each dependent variable (low back posture during crate lifting and isometric lifting strength) across each condition (baseline and instruction; lmerTest package, RStudio Version RStudio 2023.03.0, The R Foundation, Vienna, Austria). The linear mixed effects models conserved all the recorded trials (10 baseline + 3 instructions trials for each participant) by accounting for the multilevel data structure (i.e. multiple trials per participant) to avoid pseudo-replication. This statistical approach also allowed for inclusion of heterogeneity in

participant responses by estimating intercepts and slopes for each participant. The model consisted of condition (2 levels: baseline or instruction) as the fixed effect with a random slope and a random intercept for each participant. Statistical significance was considered if $p < 0.05$.

The results of the linear mixed effects models showed that individuals greatly varied in their postural and strength responses to the instruction condition (Table 7.1). This motivated an *a posteriori* analysis to classify individual responses. Changes in low back posture were first classified using the within participant trial-to-trial variation to calculate personalized confidence interval thresholds per condition (Swinton et al., 2018):

$$Confidence\ Interval\ Threshold = \sqrt{2} \times 1.96 \times SD_{WP}$$

Where SD_{WP} is the within participant trial-to-trial standard deviation. Individuals were first classified as those who would and would not benefit from reducing low back flexion based on whether their baseline condition confidence interval exceeded or was below 40% of maximum low back flexion range-of-motion, respectively. The threshold of 40% of maximum low back flexion range-of-motion marks the range of motion within which passive low back stiffness and tissue strain, which increases exponentially with low back flexion, is typically minimized (Beach et al., 2005; Fewster et al., 2021; Buchman-Pearle et al., 2023). Within each group (would and would not benefit from reducing low back flexion), low back postural responses were classified as successful (S) or unsuccessful (U) based on whether the instruction condition confidence interval was below or exceeded

40% of maximum low back flexion range-of-motion respectively. The confidence intervals were calculated from all 10 trials for the baseline condition and all 3 trials from the instruction condition.

The changes in isometric lifting strength between conditions were classified based on the combined trial-to-trial SD_{WP} across conditions. The combined variation across all trials and both conditions were used by subtracting the mean of the two isometric lifting exertion trials within each condition from each respective trial within each condition, to acquire a stronger estimate of SD_{WP} . This centered the variation within each condition at 0 by removing the difference in magnitude between conditions but maintaining the variation between the trials within each condition. Assuming that the trial-to-trial variation was consistent between conditions, the SD_{WP} for isometric lifting strength was calculated across all 4 mean bias removed maximum isometric lift trials. Individual changes in maximum isometric lifting strength were classified as an increase (I), decrease (D), or no change (S) if the difference between baseline and instruction conditions were respectively greater, lesser or within the participant specific confidence interval. The difference between conditions was calculated using the mean maximum isometric lift force achieved between each of the 2 trials for each condition.

7.3 Results

The aggregate change across conditions (i.e. fixed effect estimates) indicated that low back flexion significantly decreased ($p < 0.0001$) and isometric lifting strength significantly increased ($p = 0.018$) in the instruction condition compared to baseline. However, the linear mixed effects model estimated a very high between participant standard deviation (i.e. random effect standard deviation) of these estimated aggregate changes in isometric

lifting strength across conditions. Since the magnitude of the random effect standard deviation was larger than the fixed effect estimate, this indicates that some participants had responses that were contrary to the aggregate group results (Table 7.1; Figure 7.4B). Although the random effect standard deviation of low back posture changes was lower than the fixed effect estimate, visualizing the individual responses revealed that individuals varied in their low back postural responses as well (Figure 7.4A). Specifically, the mean change across conditions appeared to be driven by large changes in some individuals while others showed relatively small changes in low back flexion (Figure 7.4A). This motivated an *a posteriori* analysis to classify individual responses.

Table 7.1: Linear mixed effect model results for changes in low back posture and isometric lifting strength between the baseline and instruction condition. The intercept represents the value at baseline whereas the estimate is the change from baseline during the instruction condition.

	<u>Fixed Effects</u>		<u>Random Effects</u>	
	Intercept $\pm$	Estimate (β) $\pm$	SD	SD
	SE	SE	Intercept	Estimate
Relative low back flexion (%)	47.23 $\pm$ 2.49	-14.54 $\pm$ 1.97***	15.67	12.08
Isometric lifting strength (N)	603.39 $\pm$ 35.32	66.12 $\pm$ 26.76*	215.09	146.19

* $p < 0.05$, *** $p < 0.001$

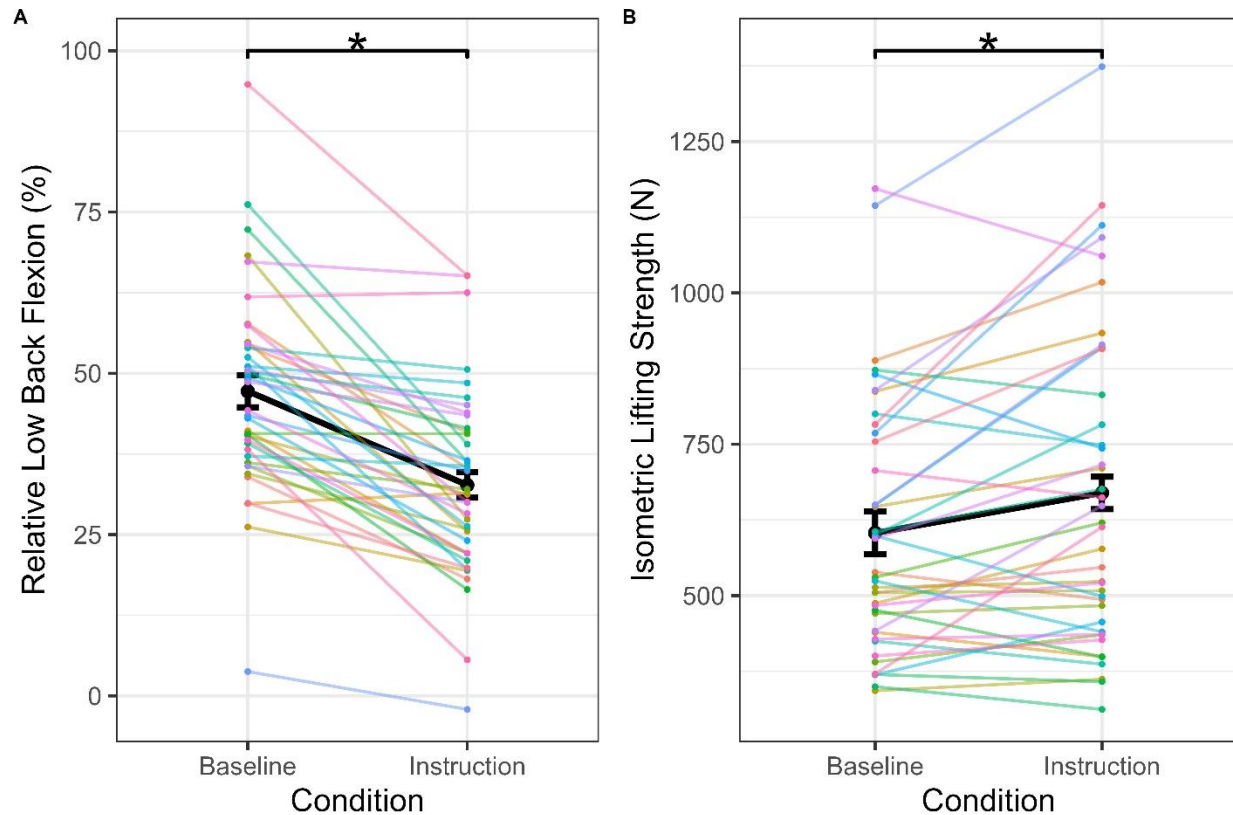


Figure 7.4: Changes in low back flexion (A) and isometric lifting strength (B) across conditions. Estimated marginal means are represented in black with the error bars representing standard error. The estimates for each participant from the linear mixed effects model are depicted by the coloured points and lines, where each colour represents a participant. Significant differences between conditions are indicated by “*”.

Individual response classification of the postures adopted during lifting is depicted in Figure 7.5. Thirty-seven participants (92.5% of total sample) were classified as individuals who could potentially benefit from a decrease low back flexion (they exhibited flexed lumbar spine postures during lifting). Within this group, 15 (37.5% of total sample) successfully decreased low back flexion to be below 40% of their flexion range-of-motion in response to the simple verbal directive whereas 22 (55% of total sample) were unsuccessful in reducing their low back flexion below this threshold. Among those that could benefit from decreasing low back flexion and successfully did so, reductions in low

back flexion ranged between 1.6 and 33.5% of maximum low back flexion range-of-motion compared to baseline (calculated as the difference between the mean of the last 3 baseline trials and the mean of the 3 instruction condition trials). Changes in low back flexion between the last 3 baseline trials and the 3 instruction condition trials among those unsuccessful in decreasing low back flexion ranged between an increase of 3.5% and a decrease of 44.8% of maximum low back flexion range-of-motion (Figure 7.5).

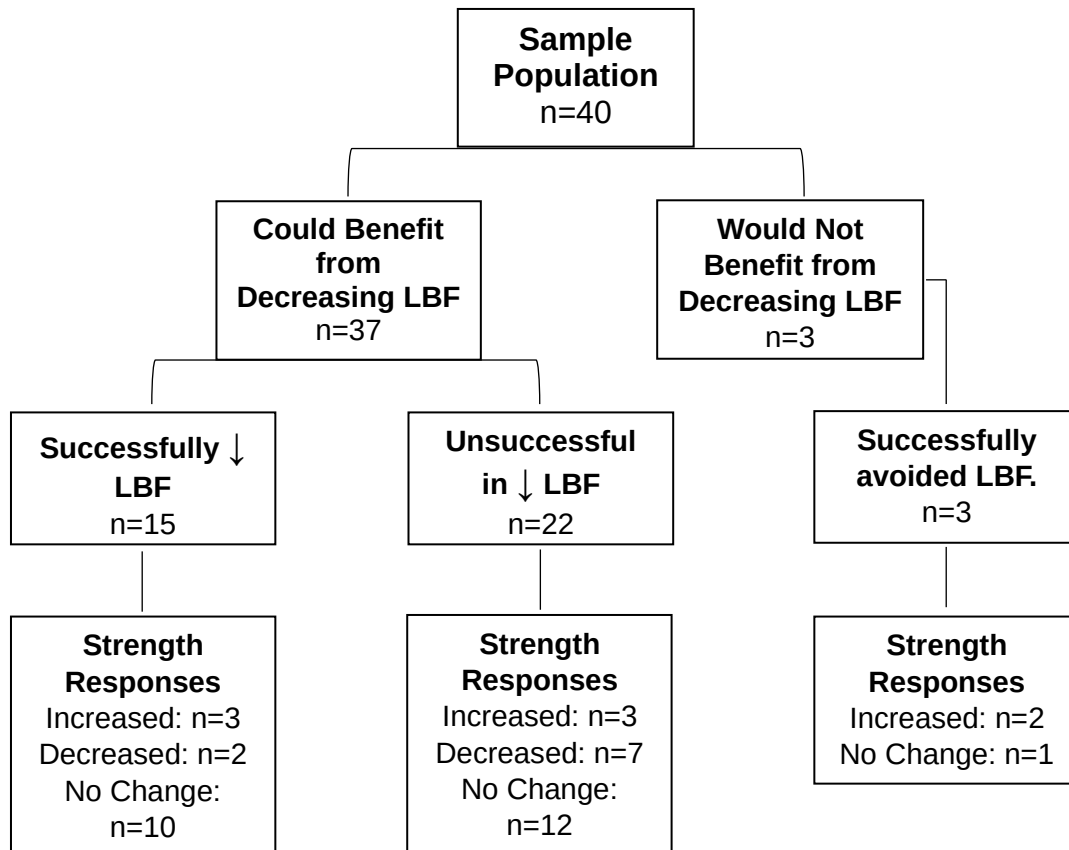


Figure 7.5: Classification of low back flexion (LBF) and isometric lifting strength responses to the directive to avoid rounding the low back.

Within the successful and unsuccessful (S and U) groups among those that were classified as benefiting from decreasing low back flexion, there were 3 types of responses in isometric lifting strength: increase (I, n=6, 15% of total sample); decrease (D, n=9, 22.5% of total sample); and no change (S, n=22, 55% of total sample). The range of increases in group I was between 106 and 449 N. The range of decreases in maximal isometric lift force in group D was between 49.9 and 212 N. The range within group S was between an increase of 224 N and decrease of 227 N. The ability to exert force was impacted among both those that were classified as successful and unsuccessful in decreasing low back flexion (Figures 7.5 and 7.6).

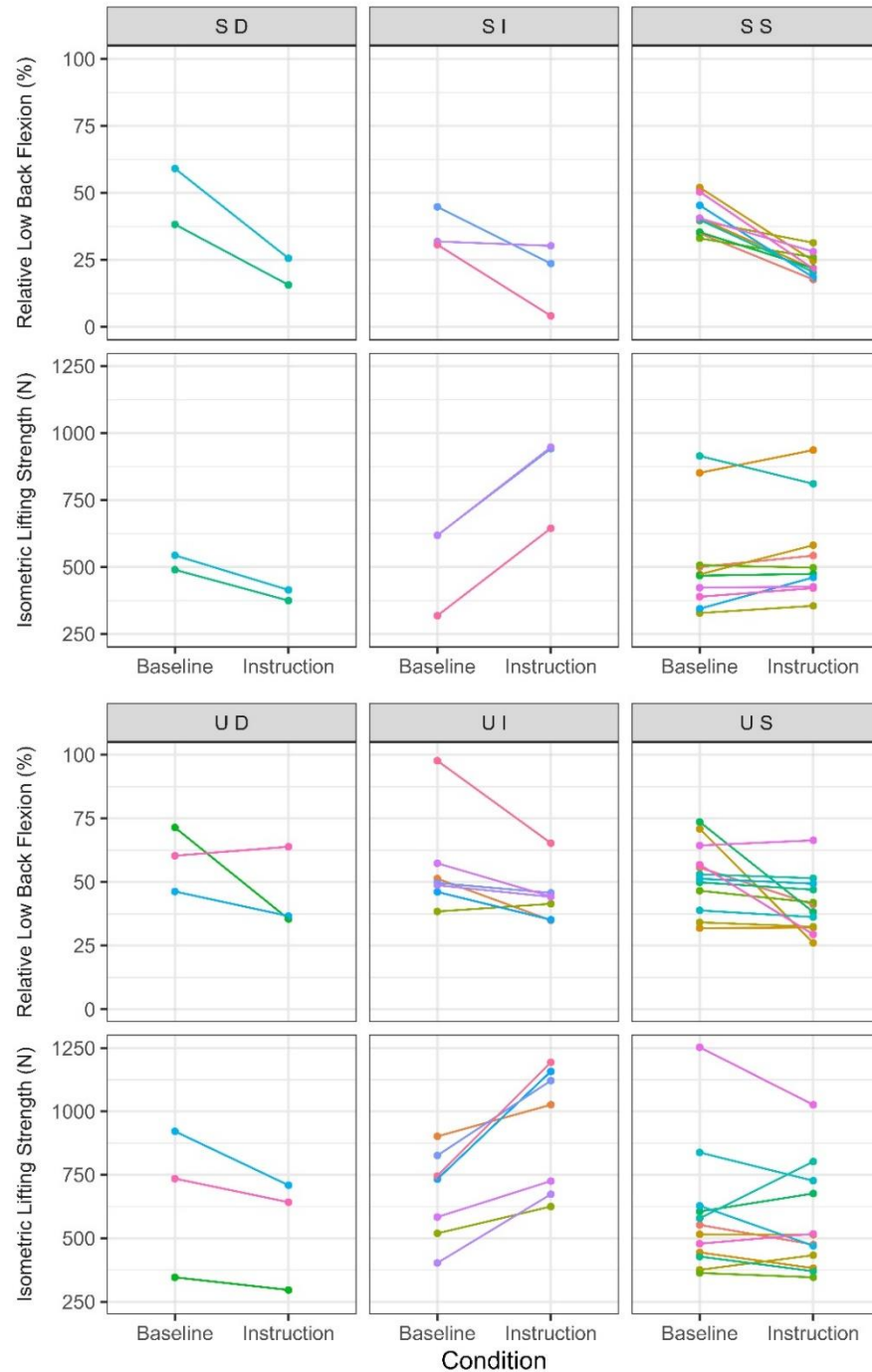


Figure 7.6: Individual responses to instruction in relative low back flexion (% maximum low back flexion range-of-motion) and isometric lifting strength among those classified as benefiting from decreasing their low back flexion. The first letter of each group corresponds to successful (S) and unsuccessful (U) low back postural responses. The second letter corresponds to isometric lifting strength changes from baseline to instruction, indicating an increase (I), decrease (D) or no change (S). Note that the changes depicted are the mean values across the last 3 baseline trials and the 3 instruction trials.

There were varying responses in hip and knee flexion/extension angle within each group such that some flexed the hip more than the knee and vice versa. This supports the hypothesis that participants utilized the greater flexibility afforded by cueing a specific key movement feature which in this case was low back posture. Those who were classified as benefiting from decreasing low back flexion but were classified as unsuccessful, exhibited changes in their hip and knee posture. This indicates that they attempted to change posture in response to the simple verbal instruction targeting the key feature of low back posture (Figure 7.7).

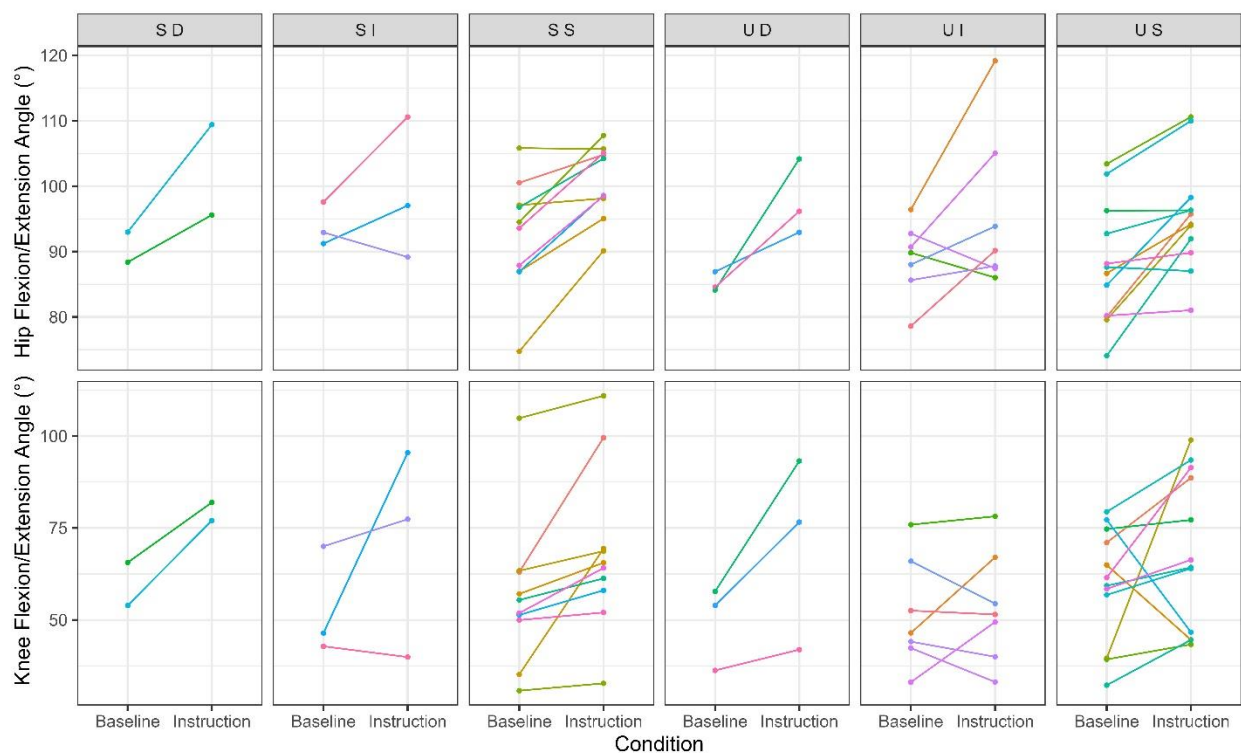


Figure 7.7: Changes in hip and knee flexion/extension angle for each response group within the group that was classified as benefiting from decreasing low back flexion. Positive angles indicate flexion and increases in magnitude indicate increases in flexion. Note that the changes depicted are the mean values across the last 3 baseline trials and the 3 instruction trials.

Three participants were classified as not benefiting from changing their low back posture as the confidence interval of their baseline trials were below 40% maximum low back flexion range-of-motion. Within this group, none of the participants increased their low back flexion such that the confidence interval crossed 40% of maximal low back flexion range-of-motion, and none of the participants decreased strength in response to the simple verbal instruction (Figures 7.5 and 7.8).

There appeared to be no association between participant sex and postural response (Successful: 9 female and 6 males, Unsuccessful: 10 females and 12 males, Would not benefit from decreasing low back flexion: 1 female and 2 males) or isometric lifting strength response (Increase: 5 females and 7 males, Decrease: 2 females and 3 males, No Change: 13 females and 10 males).

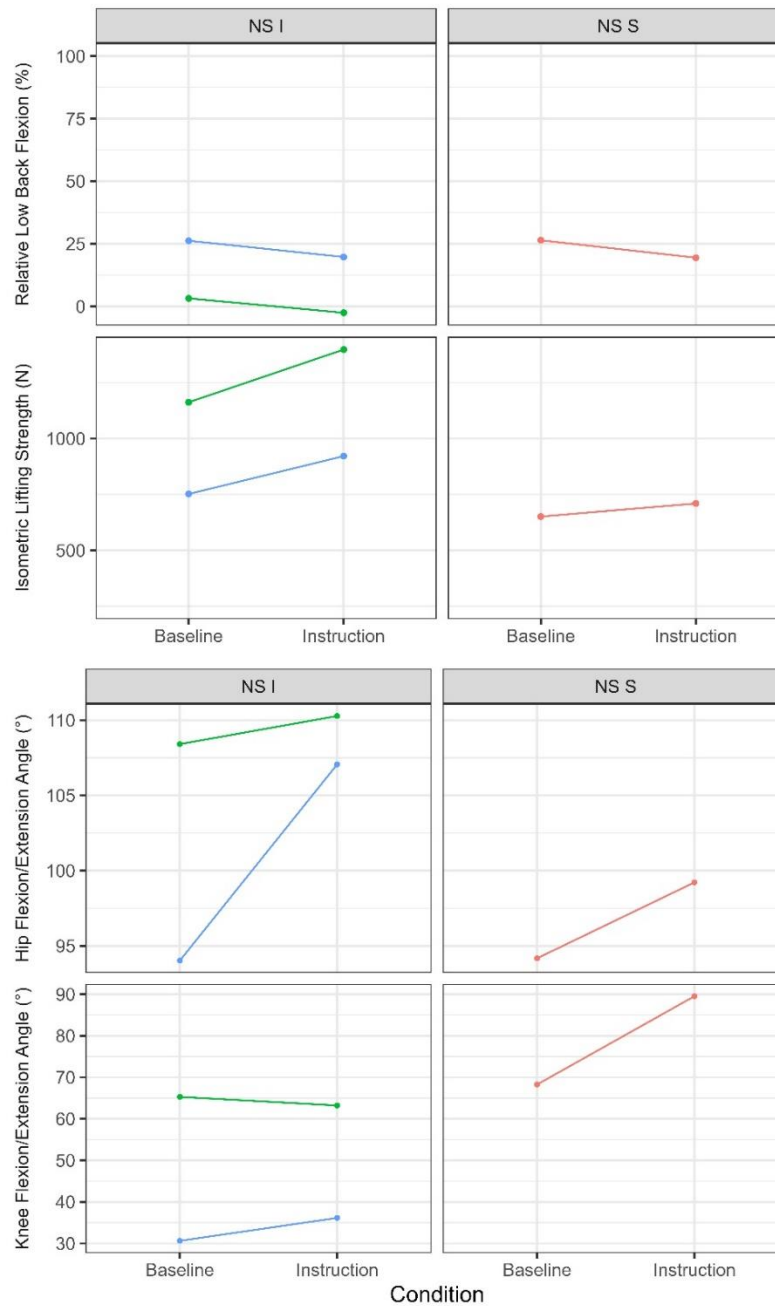


Figure 7.8: Individual responses to instruction in relative low back flexion (% maximum low back flexion range-of-motion), isometric lifting strength, hip flexion/extension angle, and knee flexion/extension angle among those classified as not benefiting from changing low back flexion. NS corresponds to those who would not benefit from changing low back posture but were successful in continuing to avoid low back flexion in the instruction condition. The second letter corresponds to isometric lifting strength changes from baseline to instruction, indicating an increase (I) or no change (S). Positive hip and knee angles indicate flexion and increases in magnitude indicate increases in flexion. Note that the changes depicted are the mean values across the last 3 baseline trials and the 3 instruction trials.

7.4 Discussion

Group-level analyses revealed that participants reduced low back flexion and increased strength after receiving the simple verbal directive to avoid rounding the low back, but there was high heterogeneity in the impact on isometric lifting strength. Although most participants (57.5% including those classified as not benefiting from changing their low back flexion) appeared to maintain their physical performance potential, there were some individuals who increased (20%) and some who decreased (22.5%) isometric lifting strength. These responses in isometric lifting strength occurred independently from changes in low back posture. This suggests that the provision of simple verbal instruction may not have the intended effect across every individual, and thus personalized interventions may be necessary to elicit the desired effect of reducing low back injury potential without limiting performance potential.

In interpreting the results, it is critical to discuss the limitations that emerged in the replication of crate lifting posture and execution of the maximum isometric lifting exertions. Many participants lost the setup position that replicated their crate lifting posture when exerting maximal force. Some of the difficulty in maintaining the setup posture could be attributed to the challenge in exerting maximal force in a posture adopted during light mass crate lifting that was not organized for maximal force exertion due to the low demand of the crate lifting task. Most participants adopted a body position where the crate was a moderate distance away from their feet. This may have potentially been due to the size of the platform that supported the crate, the area of the force platforms that participants stood on during the experiment, and the weight of the crate that may not have required participants to position their body closer to the weight. Since the handle was not fixed in

the vertical direction and could be moved, some participants would lean back with the handle or be pulled onto their toes as they exerted maximal force due to the distance between their body and the handle (Figure 7.2B), making the task awkward for them to exert maximal effort. These losses in posture appeared to be primarily balance driven rather than deviations in joint posture (e.g. change in low back curvature) which was observed during isometric lifting exertions where the handle was closer to the body (Study 1). Some of the effects on balance and issues with replicating the posture can be observed in Figure 7.2B. The difficulty in exerting maximal force in the replicated crate lifting posture may have also contributed to high trial-to-trial variability in the maximal isometric lifting trials which masked potentially practically meaningful changes in lifting strength. Since isometric lifting force responses were classified based on the trial-to-trial variability across all isometric trials collected, higher trial-to-trial variation required a higher magnitude of change for any responses to be classified as an increase or decrease. The novelty of the maximal isometric exertions may have elevated the trial-to-trial variation which led to some strength changes up to an increase of 224 N and decrease of 227 N being classified as no change (group S, Figure 7.5 and 7.6). However, these immediate changes in strength may be meaningful as they exceed changes that typically occur in deadlift 1-repetition maximum following 6 weeks of training (approximately 70 to 190 N increases depending on training experience (Androulakis-Korakakis et al., 2021; Colquhoun et al., 2018; Zweifel et al., 2017)). The current study necessitated that conditions were administered in sequence which may have allowed for better performance during the isometric exertions performed for the instruction condition by learning and adapting with each subsequent trial. Consequently, some participants

may have learned how to adjust and balance their body during the baseline condition so that they were able to apply force more effectively in the instruction condition. A large proportion (80%) of the sample participants were physically active and had previous resistance training experience involving lifting which may have facilitated learning and adapting to exert higher force with subsequent trials (Bohle Carbonell et al., 2014). However, this is difficult to determine but remains questionable due to the few isometric strength trials captured in this dataset. Further investigation is warranted to determine if addressing the abovementioned limitations would alter the results, if different sample populations exhibit different responses, and whether responses are maintained across longer time periods.

Although cueing low back posture as a specific movement feature using simple verbal instruction afforded flexibility for participants to self-organize as indicated by the varying hip and knee postural changes within each identified group (Figure 7.7), simple verbal instruction did not result in substantial decreases in lumbar flexion in over half (55%) of the participants (Figure 7.5 and 7.6) despite the group-level analyses indicating otherwise (Figure 7.4). However, it is important to consider the nuances in the method used to classify postural responses as successful and unsuccessful. The threshold used was based on previous evidence showing that passive low back bending stiffness, which can indicate strain on passive tissues (e.g. posterior spinal ligaments), is expected to be minimized below 40% of maximum low back flexion range-of-motion (Beach et al., 2005; Fewster et al., 2021; Buchman-Pearle et al., 2023). The same evidence suggests that passive stiffness exponentially increases with low back flexion such that passive tissue strain is highest beyond 80% of maximum low back flexion range-of-motion (Beach et al.,

2005; Fewster et al., 2021; Buchman-Pearle et al., 2023). In the current study, all participants exhibited less than 75% of maximum low back flexion after instruction, which due to the exponential relationship (McGill et al., 1994), should have had a meaningful effect on passive tissue strain. Yet, some of these participants were still classified as unsuccessful as the variability they exhibited during the 3 crate lifting trials in the instruction condition exceeded 40% of maximum low back flexion which marked the threshold below which passive tissue strain is expected to be minimized. It is possible that more participants would have been classified as successful with more time and practice. Additionally, it should be noted that the low and high passive stiffness thresholds mentioned in this discussion (40 and 80% of maximum low back flexion range-of-motion) can vary between individuals. Thus, the statistical thresholds used to categorize responses do not encompass all biologically meaningful changes. It may be beneficial to improve on the current individual response focused approach to better classify changes.

Keeping in mind the abovementioned methodological considerations, the current results highlight that using simple directives that target specific “key” movement features to control biomechanical loading can influence posture without hindering physical performance potential in most individuals. However, it is not guaranteed that every individual will naturally respond in a manner that maintains their physical performance potential. Strength changes were apparently independent of whether an individual was considered as successful or unsuccessful at reducing low back flexion. Due to the independence between these two outcomes, the results suggest that it is necessary to assess both an individual’s ability to achieve the targeted postural cue and their ability to do so without decreasing their physical performance potential.

Since participants utilized the afforded flexibility but not all participants did so to effectively modify low back posture as instructed, movement-based interventions may be enhanced by providing more than simple verbal directives. As opposed to simple verbal instruction, targeting key movement features using individualized movement coaching and training may provide improved efficacy in influencing movement behaviour to potentially encompass all individuals. Postural coaching and feedback has the potential to be more effective than didactic training alone (Chan et al., 2022; Kamachi et al., 2021), but responses may vary based on the type of feedback and its implementation as some type of feedback may theoretically be more successful than others (Pinto et al., 2018). Prior research has demonstrated that coaching postural features during physical exercise can positively transfer to unrehearsed non-exercise tasks (Frost et al., 2015). However, providing postural feedback and movement coaching during exercise can also result in individually varying responses (Frost et al., 2015; Pinto et al., 2018), requiring knowledge and strategies to assess, coach, and train on an individual basis. Personalized coaching and training strategies may equip individuals with the competencies and capacities to coordinate and control their movements in ways that attenuate biomechanical loading while simultaneously satisfying task objectives without hindering physical performance. This aligns with recent rationale to equip workers with the capabilities, opportunities, and motivations to self-monitor and -regulate their movement rather than promoting one-size-fits-all directives for safe movement such as the squat lift technique (Denis et al., 2020). The results from the current study highlight that these personalized assessment and coaching strategies should encompass both biomechanical loading and physical

performance potential as changes in isometric lifting strength were independent from changes in low back posture.

7.5 Conclusion:

Using simple verbal instruction targeting specific movement features to regulate biomechanical loading can influence posture without hindering performance potential in most individuals. However, some individuals can respond in a manner that decreases performance potential and may require more detailed movement coaching and training to modify movement behaviour without hindering physical performance potential.

Chapter 8 : General Discussion

8.1 Thesis Contributions

The global aim of this thesis was to investigate how posture and trunk muscle coordination can influence the ability to exert force during lifting. Study 1 of this thesis demonstrated that modifying posture can impact isometric lifting strength, but the effect magnitude and direction (increase or decrease) varied greatly across individuals (Chapter 4). Study 2 demonstrated that cueing trunk muscle coordination can influence the ability to exert maximal isometric lifting force. Results indicated that trunk muscle coordination should not be cued if physical performance is the immediate priority. Although the results did not clearly elucidate how trunk muscle coordination can influence the ability to exert end-effector force, the results informed the design of future research to further investigate the effects of trunk muscle coordination on the ability to exert force in multi-joint tasks (Chapter 5). In the interest of identifying quantification approaches that could be used to capture and estimate the individually varying effects of posture on strength, Study 3 sought to compare kinematic-based proxies for leverage that were used in prior research but not verified to capture leverage in multi-joint tasks (Chapter 6). The results demonstrated that the investigated kinematic-based proxies for leverage are not best suited for assessing the individual effects of posture on the ability to exert isometric force in multi-joint tasks. Study 4 indicated that most individuals maintain their physical performance potential when given postural cues during low external demand tasks that do not acutely require high levels of physical exertion. However, some individuals may not self-organize their posture in a manner that preserves the ability to exert force in

response to simple verbal directives targeting low back posture. These individuals may require specific movement coaching and training.

8.1.1 Specific Research Contributions

Study1: The Effects of Trunk Inclination and Low Back Curvature on Lifting Strength.

In Chapter 4, the effects of trunk inclination and low back curvature on the ability to exert maximal force were dissociated by experimentally and statistically controlling for trunk inclination and low back curvature. Participants performed 4 controlled postural conditions that included 2 trunk inclinations (vertical and horizontal) and 2 low back curvatures (neutral and flexed). The hypothesis that modifying trunk inclination would affect more joints and muscles to have a greater impact on isometric lifting strength compared to low back curvature was rejected as each of these postural characteristics showed a potential to affect maximal isometric lifting force to a similar extent. However, the results confirmed the hypothesis that the effects of trunk inclination and low back curvature on isometric lifting strength depend on one another. Specifically, the effects of modifying posture on strength in multi-joint tasks can vary greatly across individuals in magnitude as well as direction (increase/decrease in force exerted). Findings also suggest that strength in multi-joint tasks cannot be inferred solely from observing posture and that a single posture cannot be generalized as being the strongest for every individual. These results question previous conclusions drawn from aggregate group analyses that promote flexing the low back to increase lifting strength (Mawston et al., 2021). Additionally, the results contrast the notion that there is always a trade-off between physical performance and biomechanical demand as it is not necessary that these two objectives are always in opposition. The heterogeneity in results suggest that the relationship between posture's

effects on biomechanical demand and physical performance may be modifiable through training, such that individuals can adapt to being stronger in postures that are also favourable for tissue loading and tolerance. In fact, the preliminary analysis on L4/L5 joint contact and compression shear forces indicate that this should be the target for training as the cost of flexing the low back on joint loading and tolerance is much higher than potential gains in acute physical performance potential.

Study 2: The Effects of Cueing Trunk Muscle Coordination on Isometric Lifting Strength.

In Chapter 5, the effects of cueing trunk muscle coordination using simple verbal directives and detailed coaching that included verbal and physical cueing were investigated. The simple verbal directive was “activate your core” which is a common directive in applied settings and akin to the simple directives described to cue trunk muscle coordination in previous research. The coached condition included both verbal and physical cues which have been promoted in applied settings but not yet been thoroughly investigated. Result confirmed the hypothesis that detailed coaching would be more effective than simple directives at creating muscular coordination that stiffens the spine to reduce low back postural displacement. However, the immediate response to both simple directives and detailed coaching reduced isometric lifting strength. The quantitative results coupled with the qualitative participant experiences that were recorded, suggest that a detailed coaching approach can be more effective and consistent at modifying trunk muscle coordination to increase spine stiffness. The qualitative results also helped identify that investigating the immediate effects of modifying trunk muscle coordination may mask the full potential of the effects on posture

and strength due to the imposed cognitive demand. Given that many studies have only observed the immediate effects of modifying trunk muscle coordination to investigate its impact on biomechanics and physical performance (Brown et al., 2006; Coenen et al., 2017; Kavcic et al., 2004; Liebson et al., 2009; Norrie & Brown, 2020; Reeves et al., 2006; Stanton & Kawchuk, 2008; Vera-Garcia et al., 2006, 2007), this is an important finding that informs the interpretation of previous evidence and suggests longer-term study designs for further investigation. Longer-term study designs would elucidate how cueing trunk muscle coordination may influence motor execution and performance without the confounding effects of cognitive demand and attentional focus that would be reduced with practice and learning.

Study 3: Explaining the Individually Varying Effects of Posture on Isometric Lifting Strength.

Motivated by the results from Study 1 (Chapter 4), Study 3 presented in Chapter 6 aimed to evaluate the extent to which proxies for leverages derived from kinematic data could explain the individually varying effects of posture on the ability to exert force. These kinematic-based proxies included the joint external moment arms and predicted joint-angle-dependent torque-generating capacities, which have been previously used in research to describe the effects of posture on the ability to exert force in multi-joint tasks. However, these kinematic quantities have not been thoroughly verified to capture the individual variability in the ability to exert force in multi-joint tasks. Understanding the extent to which such quantities could capture and explain individual variability in the effects of posture on the potential to exert force, could facilitate the development of a

quantitative approaches to assess the effects of posture on physical performance without needing direct measurements of physical performance. Additionally, if these kinematic quantities could reasonably describe leverage, they could be coupled with postural assessments of biomechanical demand to make inferences in practical settings. As hypothesized, the external moment arms and predicted torque capacity which are derived from kinematic quantities, were not able to explain the variation in the effects of posture on isometric lifting strength on an individual basis. In contrast and as expected based on fundamental biomechanical principles, the net joint reaction moment explained a high proportion of the total variation in the effects of posture on isometric lifting strength without needing to independently account for any individual differences in the effect of the net joint reaction moments on isometric lifting strength. Thus, both kinematic and kinetic measurements may be required to assess the effects of movement on physical performance and subsequently modify movement to enhance physical performance.

Study 4: Impact of Individual Responses to Postural Directives on Isometric Lifting Strength.

In chapter 7, the extent to which instruction to avoid rounding the low back (i.e. lumbar spine flexion) during light mass lifting could immediately impact isometric lifting strength was investigated. The natural postural responses to the verbal low back postural directive during low mass lifting were replicated during maximal isometric lifting exertions, to determine the impact of simple verbal directives can have on physical performance potential in a low external demand task. Although aggregate changes indicated that instruction decreased low back flexion and increased isometric lifting strength, there was

high heterogeneity which again motivated examining individual responses. Thirty-seven participants exceeded 40% of their low back flexion range-of-motion during the baseline condition and were classified as potentially benefiting from reducing their low back flexion. Among this group, 15 and 22 participants were respectively classified as successful and unsuccessful in decreasing low back flexion below the threshold of 40% of flexion range-of-motion which marked the range-of-motion within which passive low back stiffness and tissue strain, which increases exponentially with low back flexion, is typically minimized. Though replicating the crate lifting posture to assess isometric strength emerged to be challenging, there were individuals who increased ($n=6$), decreased ($n=9$) and did not change ($n=22$) isometric lifting strength, independent of their low back postural response. While most individuals appear to maintain physical performance potential when given simple verbal directives targeting specific movement features to regulate biomechanical demand, it is not guaranteed that every individual will do so. This suggests that the provision of simple verbal instruction may not have the intended effect across every individual, and thus personalized interventions may be necessary to elicit the desired effect of reducing low back injury potential without limiting performance potential.

8.2 Thesis Limitations

The following section includes limitations to the thesis beyond what has been already discussed in the chapters describing each of the 4 studies performed.

8.2.1 Single Data Collection

A limitation of this thesis is that the data used to answer the research questions were collected within a single session. At a global level, this limited the flexibility to design and

modify research objectives based on the initial findings from Study 1 and Study 2. Although the analyses of the data attempted to account for the effects of fatigue where possible, results may have been influenced by the duration of the data collection session which lasted between 4.5 to 6 hours. The procedures also demanded multiple high effort exertions, and it is unclear to what extent the results were influenced by the number of maximal efforts performed. Previous studies have investigated 56 (Kumar & Garand, 1992) and 90 (Chen et al., 2011) maximum isometric lifting exertions and 8 (Beckham et al., 2018), 12 (Guppy et al., 2019), and 18 (Comfort et al., 2015) maximal isometric mid-thigh pulls within a single data collection session, but did not report observing fatigue. Participants were frequently asked about their fatigue and although some reported being slightly fatigued compared to the start of the data collection, none felt like it influenced their strength within each experimental block. It would have been beneficial to track fatigue throughout the data collection or to perform an experiment to determine the consistency of multiple maximal isometric lifting exertions. Baseline trials were collected for each experimental block/study. Thus, comparisons were made to baseline and physical states relatively close in time to one another.

It would also be of interest to understand the necessity for familiarization with the isometric exertion as it may have been a novel task for many. Participants were provided a warmup that included isometric exertions at submaximal efforts, but it is unclear if and how much practice might modify the current results.

8.2.2 Isometric Lifting Exertions

The studies in this thesis focused on isometric exertions in a position emulating initiation of a lift from the ground. Isometrics were selected to facilitate more controlled modification

of posture and muscle coordination while reducing the accumulation of fatigue. It was expected that using dynamic efforts would make it difficult to maintain desired postures and muscle coordination which could have obscured the true effect of changing an individual's movement on physical performance. For instance, previous practice as well as the physical and cognitive demands imposed by maintaining the targeted posture or muscle activity during maximal dynamic exertions could have resulted in an individual not maintaining the desired movement profile or executing the movement with compensations (Farrow & Robertson, 2017; Pinto & Dickerson, 2021). This would not have reflected how the change would actually influence physical performance without providing sufficient time and practice to become equally proficient with each postural or muscle coordination condition. Additionally, becoming equally proficient may not be possible. Interestingly, some of these issues were still experienced with the isometric strength test. Participants lost the targeted joint posture during the isometric exertions in Study 1 (Chapter 4) and many participants noted it was difficult to maintain the cued muscle coordination while performing maximal efforts in Study 2 (Chapter 5). Although the isometric exertions allowed for crate lifting posture to be replicated to test strength of individual postural responses in Study 4 (Chapter 7), the replicated posture was challenging to maintain and appeared to affect how participants positioned the system centre of mass relative to its base of support during the exertion.

The implications of changes in isometric strength on dynamic activities would require consideration of other biomechanical factors such as mechanical work. For example, selecting a position that maximizes an individual's isometric lifting strength may not necessarily impact dynamic strength. Isometric strength testing may only capture a

subset of factors that contribute to dynamic strength (James et al., 2024). However, there is evidence that suggests isometric lifting strength tests are generally associated with dynamic physical performance in multiple tasks (Lum et al., 2020). Future work should aim to distinguish between biomechanical features captured by isometric testing and how they may impact dynamic physical performance to facilitate inferences from isometric testing.

8.3 Global Thesis Conclusion

Given the scarcity of knowledge about how posture and trunk muscle coordination can influence the ability to exert force in multi-joint tasks such as lifting from the ground, this thesis has made several novel contributions. Results demonstrate that modifying posture and trunk muscle coordination have a substantial potential to influence the ability to exert force in multi-joint tasks and thus the effect on physical performance cannot be ignored during movement assessments and interventions that target reductions in biomechanical demand. The findings from this thesis also inform practices to enhance physical performance and development in conjunction with limiting unnecessary biomechanical demands that could increase risk of LBD. In contrast to modifying posture and trunk muscle coordination to influence biomechanical demand and contrary to previous notions, the effects of modifying movement features such as posture and trunk muscle coordination on the ability to exert force at the end-effector are more complex than initially hypothesized based on previous evidence. Specifically, the effect of posture on the ability to exert isometric force in multi-joint tasks can vary greatly across individuals and cannot be inferred from observing posture or proxies for leverage derived solely from kinematic data as indicated by Studies 1 and 3 (Chapters 4 and 6). Contrary to previous notions

that peak force exertion at the end-effector during lifting requires adopting low back flexion which imposes a higher risk of developing LBD (Mawston et al., 2021), it is not necessary that maximizing the ability to exert lifting force requires adopting flexed low back postures. This is because a single postural profile cannot be universally accepted as the strongest for every individual during multi-joint tasks, as indicated from the results of Study 1 (Chapter 4) that showed that flexing the lower back can increase, decrease or not change lifting force exerted at the end-effector across individuals. However, results from Study 1 (Chapter 4) also indicated that the effect of low back flexion on the ability to exert lifting force varies with trunk inclination which resulted in the identification of 7 different response groups. As opposed to the paradox that peak physical performance requires postures that impose higher than necessary biomechanical demands, the individual variability in the effect of posture on the ability to exert force suggests that individuals can potentially adapt with training to be stronger in postures that favour reduced tissue loading and increased joint failure tolerance. Study 1 (Chapter 4) also demonstrated the cost of adopting a flexed low back posture which imposes higher lumbar spine joint compression and shear forces, outweighs the benefits that may potentially be acutely gained in the ability to exert isometric lifting force. Together this evidence supports that low back posture should be assessed especially during high demand physical activities and that movement interventions should target attenuation of high magnitudes of low back flexion during high demand physical activities.

This thesis demonstrated that interventions to modify low back postural behaviour may require more than simple verbal directives as indicated by the results from Study 4 (Chapter 7) which suggests that most individuals may need additional movement

coaching and training to reduce their low back flexion within a range that minimizes passive tissue strain and potential injury risk (within 40% of their maximum flexion range-of-motion). Specifically, 22 of the 37 participants that exhibited more than 40% of their maximum low back flexion during baseline condition crate lifting, were not able to reduce their low back flexion below 40% of their maximum flexion range-of-motion with a simple verbal directive to “avoid rounding your low back”. Study 4 (Chapter 7) also provided new evidence that specifically cueing low back posture as a “key feature” (i.e. in a manner that affords flexibility to organize the posture of the rest of the body linkage) during a task with low external demands, does not negatively affect physical performance potential for most individuals. This was indicated by the maintenance of isometric lifting strength of the postures exhibited during light mass crate lifting at baseline and in response to the directive. However, Study 4 (Chapter 7) also showed that some individuals do not maintain their physical performance potential even when afforded the flexibility to self-organize the posture of the rest of their body while attempting to reduce low back flexion, as they adopted postures that decreased isometric lifting strength. For these individuals especially, it becomes necessary to develop strategies to assess and modify movement to enhance physical performance potential. Yet, this requires the development of methods that are sensitive to individual differences.

Although kinematic-based proxies for leverage such as the joint-to-external force moment arms and predictions of joint torque-generating capacity based on normative data, may be accessible and easier to conceptualize in applied settings, Study 3 (Chapter 6) showed that they do not capture the effects of posture on the ability to exert isometric force in a multi-joint task on an individual basis. Instead, results from Study 3 (Chapter 6)

suggest that quantitative movement assessment of an individual's leverage may best be described by variables derived from both kinematic and kinetic quantities such as the net joint reaction moments calculated using a rigid linked segment model. This informs the development of quantitative methods to assess leverage in multi-joint tasks which will enable an assessment of physical performance potential alongside biomechanical demand to enable improved and more comprehensive assessments and interventions that consider both biomechanical demand and physical performance.

With regards to trunk muscle coordination, Study 2 (Chapter 5) demonstrated that attempting to acutely modify trunk muscle coordination immediately interferes with the ability to exhibit maximal isometric force at the end-effector. This thesis was the first to identify the differences between using simple verbal directives and detailed coaching in modifying trunk muscle coordination. Study 2 (Chapter 5) demonstrated that the detailed coaching approach was more effective than simple verbal directive to “activate your core” in increasing muscle co-contraction and maintaining low back posture during maximal isometric lifting exertions. An important contribution from Study 2 (Chapter 5) is that immediate responses to cueing trunk muscle coordination are insufficient in capturing the full potential of the effects of modifying trunk muscle coordination on physical performance. This is a key contribution as a majority of previous research has focused on immediate responses and short-term adaptations to modifying trunk muscle coordination using simple verbal directives which may mask the full potential of any possible effects on physical performance. Future work would need to design experiments to provide time to learn and practice new trunk muscle coordination patterns to completely evaluate the effects of muscle coordination on the ability to exert force at the end effector.

The findings from this thesis inform movement coaching across multiple domains. The varying effects of low back curvature on isometric lifting strength suggests that low back flexion beyond a range that can be considered neutral, is not required to maximize strength in multi-joint tasks. The varying effects also suggest that individuals may adapt in response to training to be stronger in more neutral low back curvatures to reduce the risk of developing LBD. In fact, the corresponding estimated cost in biomechanical demand suggests that high magnitudes of low back flexion should be avoided during high demand activities such as resistance exercise for work, sport and health. Movement coaching during exercise or ergonomics should target low back curvature as a key postural feature instead of targeting vaguely defined whole body postural techniques (e.g. stoop and squat lift or sumo and conventional deadlift) to influence low back biomechanical demand. This focus on lumbar curvature is important as the increase in biomechanical demand with low back flexion is much higher compared to modifying the rest of the body linkage to alter trunk inclination. Additionally, targeting low back curvature as a key feature has the potential to allow individuals to self-organize their movement without hindering physical performance compared to providing strict instructional constraints on whole body posture which can hinder performance and not be suitable for every context or task. However, the results also highlight that solely providing simple verbal directives are insufficient in modifying movement features such as posture or trunk muscle coordination, suggesting the need to develop more effective coaching strategies. Movement coaches and researchers should attend to whether the cues used are effective in eliciting the desired outcome. The effects of posture on the ability to exert force that were observed in this thesis also inform physical capacity testing for work, sport and

health. Strength tests may not always indicate physical performance potential of tasks that require very different postures and posture is important to consider when interpreting tests. Although kinematic measurements are more accessible and easier to observe and interpret, inferences about the ability to exert force on an individual basis should not be made by solely observing posture or using simple kinematic based proxies such as the joint-to-external force horizontal distances.

Overall, this thesis uncovered the complexity in the effects of posture and trunk muscle activity on the ability to exert force at the end-effector during an exemplary multi-joint task. The complexity observed motivates and informs further investigation to develop improved strategies assess and enhance physical performance while regulating biomechanical demand.

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