A COMPARISON OF METHODS TO QUANTIFY CONTROL OF THE SPINE

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A thesis submitted in partial fulfillment of the requirements for the
M.Sc. degree in Human Kinetics

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Acknowledgements

Despite my name being on the first page of this document, this project is very much a collaboration with many great, intelligent people that I would like to thank. I feel as though this thesis has created more questions than answers for me, and the only thing that I can say with complete confidence is that I could not have done this on my own.

First and foremost, I would like to thank my supervisor, Dr. Ryan Graham. The opportunities to learn that you have provided me with are invaluable. I truly believe the research and learning environment that you have created is the gold standard, and the fundamental principles that are the foundation of our lab will stay with me for life.

Next, I would like to extend my appreciation to Dr. Jaap van Dieën, Dr. Julie Nantel and Dr. Kristi Adamo. The time that you all committed to my project out of your busy schedules improved this thesis immensely. I hope to be able to pay it forward and assist someone with the same dedication that you have all demonstrated to me.

Third, I would like to thank my lab mates. This journey would have been much more difficult without the contribution of every single one of you. It has been a pleasure to develop as researchers together, and I will treasure every moment we had. Matthew Mavor, you deserve specific recognition. Since Nipissing University, you have provided answers to my endless questions and I am forever grateful for your assistance and friendship.

Finally, I would like to thank my family. Mom, Dad, Greg, and Bianca: you have all created a safe and happy home for me away from school. I am positive that I would have lost my sanity throughout this project if it were not for the support, in all its forms, from every one of you. Although there may be days it is unsaid, please remember that I love you all and I cherish the relationship I have with every one of you. From the bottom of my heart, thank you!

List of Abbreviations

λ_{max}	Maximum finite-time Lyapunov exponent
3D	Three-dimensional
6D	Six-dimensional
12D	Twelve-dimensional
ANOVA	Analysis of variance
B	Lumbar damping
Comp	Complex
COP	Centre of pressure
DV	Dependent variable
FE	Flexion-extension
FRF	Frequency response function
IV	Independent variable
K	Lumbar intrinsic stiffness
K_a	Muscle spindle acceleration feedback gains
K_p	Muscle spindle position feedback gains
K_v	Muscle spindle velocity feedback gains
LBP	Low back pain
LDS	Local dynamic stability
MS	Muscle spindle
NMC	Neuromuscular control model
PC	Pain catastrophizing
Rot	Rotation
SD	Standard deviation
sEMG	Surface electromyography
SI	Systems identification
T_{10}	Tenth thoracic vertebrae
T_{12}	Twelfth thoracic vertebrae
VIF	Variance inflation factor

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ABSTRACT

Low back pain (LBP) affects many individuals worldwide. The established association between LBP and spine motor control has led to the development of many control assessment techniques. To understand the association between motor control and LBP, it is essential to understand the relationship between separate assessment techniques. Systems identification (SI) and local dynamic stability (LDS) are two methods commonly used to quantify spine control. SI provides a detailed description of control but uses linear assumptions, whereas LDS provides a "black box" non-linear assessment and can be quantified during dynamic movements. Although both SI and LDS techniques aim to measure the control of the spine, each employs different experimental setups and data processing strategies. Therefore, the purpose of this thesis was to compare the motor behaviour outcomes of SI and LDS quantification techniques. To do this, 15 participants completed two tasks (SI and LDS) in a random order. For the SI task, participants were seated and ventrally perturbed at the level of the 10th thoracic vertebrae (T_{10}). They completed this task under instructions to resist the perturbations (resist condition) or relax and remain upright (relax condition). Admittance was represented using frequency response functions, and a validated neuromuscular control model quantified lumbar stiffness, damping and muscle spindle feedback gains. The LDS task involved participants completing three repetitive movement blocks consisting of flexion/extension, axial rotation, and complex movements. In each block, the maximum finite-time Lyapunov exponent (λ_{max}) was estimated. A stepwise linear regression determined that λ_{max} during the rotation task was best predicted by SI outcomes in the relax condition (adjusted R square = 0.65). Many conditions demonstrated no significant relationship between λ_{max} and SI outcomes. These findings outline the importance of a consistent framework for the assessment of spine control. This could improve clinical assessment efficiency as well as the understanding of the association between LBP and motor control.

CHAPTER 1: INTRODUCTION

1.1 Rationale for research

Low back pain (LBP) is a prevalent issue in today's society with a reported 70-85% of individuals experiencing LBP in their lifetime, along with an average point prevalence of 30% (Andersson, 1999). Consequently, this places a large economic burden on the health care system. In Canada, long-term disability from back and spine injury accounts for 4.7 billion dollars of the total 12.6 billion associated with all musculoskeletal injuries (Health Canada, 2002). This is the largest percentage of all musculoskeletal injuries even when arthritis and injuries of the lower limbs, upper limbs and hips are considered. Medical expenditures alone for LBP are estimated to cost between $6 and $12 billion dollars annually in Canada (Bone and Joint Canada, 2014). These associated costs are in addition to loss of productivity and disability payments. Despite the magnitude of this issue, the causes of LBP are largely misunderstood. Evidence of this is reflected in diagnosis, as 90% of cases cannot be attributed to a known pathology and are therefore labeled "non-specific" (Krismer & van Tulder, 2007). As understanding the problem is essential to finding a solution, there has been a considerable amount of research dedicated to identifying risk factors associated with the development, progression and recurrence of LBP.

Currently, it is well established that the cause of LBP is multifactorial and the biopsychosocial model has been proposed as an all-encompassing approach that accounts for: anatomical and physical aspects of health, psychological characteristics and social factors. Specifically, anatomical and physical aspects of health encompass pain-causing tissue pathologies that relate to disc disorders, nerve damage, and skeletal muscle damage (Petersen et al., 2003). Psychological characteristics of the model include aspects such as pain-related fear, which may subsequently cause individuals to adapt movement avoidance behaviour, physical activity avoidance and disability (Bunzli, Smith, Schütze, Lin, & O'Sullivan, 2017). Movement avoidance can also be detrimental mechanically, as one subgroup of LBP may stiffen the spine to limit range of movement, which can produce increased compressive loads on tissues (van Dieën, Reeves, Kawchuk, van Dillen, & Hodges, 2018). Social factors included in the model are related to aspects of job satisfaction and the work environment, which are also associated with LBP (Bigos et al., 1991); however, the exact mechanisms of this association are not well understood. While all of these factors may influence LBP risk, the degree to which these factors affect LBP is the target of current research.

When considering physical aspects of LBP, spine motor control has proven to be a promising focus. Overall, motor control is a broad term that describes how a desired posture or movement is achieved. The spine system can be divided into three main subsystems: active, passive, and neural. The active subsystem consists of active muscular properties, the passive subsystem considers muscle viscoelasticity and contribution from tendons, ligaments and bones, while the neural subsystem includes sensory feedback and neural control centres (Panjabi, 1992). It is the complex interaction between the subsystems that achieves a desired spine state. Adding to this complexity, is that contributions to spine control from each subsystem can change depending on task context. This is observed in the flexion-relaxation phenomenon, where at full spine flexion, most forces are distributed onto passive tissues, and muscular (i.e. active) contribution is reduced almost completely (McGill & Kippers, 1994). The complexity of this system can also be detrimental, because although dysfunction in one of the subsystems can be compensated for in another, this compensation may require the recruitment of movement patterns that further degenerate other systems. This mechanism could occur within the spine if, for instance, there is a degradation of passive tissues resulting in reduced force contribution from these structures. In order for the spine to meet stability requirements, stiffness contribution by additional recruitment of active tissues (and co-contraction) could lead to accelerated levels of muscular fatigue and spinal compression loads (van Dieën & De Looze, 1999).

It is well established that LBP is associated with changes in neuromuscular control, suggesting that it may interfere with one or more of the subsystems. The control outcomes resulting from LBP are debated within the literature, with some findings demonstrating increased trunk stiffness during unstable sitting (Freddolini, Strike, & Lee, 2014) while others suggest impaired control due to delayed muscle response times and balance performance (Radebold, Cholewicki, Polzhofer, & Greene, 2001). These discrepancies could be due to subgroups existing within the LBP population; however, they could also be due to different methodological approaches being used to quantify motor control. Without consistent quantification techniques or methods that classify movement behaviour similarly, exact neuromuscular control differences can be difficult to detect. This can lead to confusion within the literature and may lead to misinterpretation of the cause/effect role of spine motor control on LBP.

Despite the potential confusion that may arise from having different methods of quantifying motor control of the spine, there are advantages that are associated with diversity of outputs. For instance, systems identification (SI) techniques can be used to quantify spine neuromuscular control and identify the magnitude of contribution of different subsystems (i.e. active, passive, and neural) to maintenance of a static, upright posture. Although this method works under the assumption that the spine system is generally linear, we can obtain highly detailed results. It is also limited to the response during upright postures and it is unknown whether results can be applied to dynamic movements where neuromuscular demands may change. Local dynamic stability (LDS) on the other hand allows non-linear dynamic control to be quantified during repetitive spine movement in all planes; however, it is unable to identify the specific mechanisms that are contributing to the gain/loss of control. In addition to different outputs, SI and LDS also have drastically different methods of collecting the required data. SI requires specialized, cumbersome equipment such as a magnetically driven linear actuator, contact probe, and surface electromyography (sEMG) sensors. Additionally, SI requires the use of customized software to collect the contact force, actuator displacement and sEMG data during the perturbation protocol. On the other hand, LDS only requires the use of a single data series such as the lumbar spine Euler angles or linear and angular velocities, both of which can be obtained from small wearable sensors. Although there are differences in collection techniques and outputs, both methods capture the ability of the spine to respond to perturbations, which are experienced regularly during human movement. These perturbations can arise from the environment or error within the system itself. Environmental, or external, perturbations can be caused through the interaction of an individual with their surroundings, such as moving unstable loads or walking on slippery floors (Beaudette, Graham, & Brown, 2014). External perturbations can also be applied mechanically to systems in a research setting to observe the response and obtain information regarding that system. Local, or internal, perturbations occur within the human neuromuscular control system due to error (overcompensation) or internal disturbances such as breathing. Despite SI and LDS being utilized during different tasks, SI observes the spine system response to mechanical perturbations (i.e. rapid loading) and LDS quantifies the movement response to local (i.e. neuromuscular) perturbations.

When considering the advantages and disadvantages of SI and LDS, it is clear that these methods are complimentary; SI requires cumbersome and highly specialized equipment and software and is therefore impractical for clinical assessment of control; however, LDS may be used to provide clinical movement screening due to its accessibility. Further, SI provides extensive detail into exact mechanisms of postural control, whereas LDS does not. This can be advantageous as LDS assessment could be used in clinics with simple equipment and software, and a patient could be sent for further analysis using SI techniques if they demonstrate poor movement behaviour in the initial test. Conversely, if good movement control is observed during the LDS testing, then other factors for the onset and persistence of LBP could be explored more heavily to target the exact cause of LBP in the individual. This is increasingly important when considering the multiple factors that affect LBP and the resulting individuality of each case. By implementing effective assessment strategies, the cause, and subsequently the solution, of LBP cases may be clearer.

Although LDS and SI could effectively work together, this system of motor control assessment cannot be confidently utilized as the relationship between the motor control outcomes of SI and LDS is currently unknown. If no relationship exists between the outcomes of SI and LDS, clinical screening using LDS may be meaningless if SI were to be used as a follow-up assessment to determine the contribution of different systems. Therefore, the purpose of this thesis is to understand the relationship between the outputs of SI and LDS to: i) improve the understanding of motor control assessment strategies and ii) recognize if these techniques can be used in concert to improve control assessment in clinics. When considering that the human spine system reacts similarly to local and environmental perturbations (Mavor & Graham, 2015), it was expected that these two techniques will quantify similar control behaviour. Further, to successfully complete SI and LDS tasks, constant stabilization of the low back is required. Therefore, there is likely shared control strategies between the two tasks. Under different task instructions, SI can be used to quantify both a maximum and natural level of postural control. A maximum level of control can be defined as the control demonstrated when an individual is consciously trying to remain as stiff as possible, whereas a natural level of control is demonstrated when no specific control strategy is prescribed. As there are no maximum control instructions during the LDS task, it was hypothesized that the strongest relationship will exist between SI outcomes under natural task instructions and LDS outputs. As LDS has been applied as a spine control assessment technique during repetitive movement in all movement planes, it was hypothesized that repetitive flexion and extension will exhibit the closest relationship with SI outcomes as both tasks involve movement in the anterior/posterior direction.

CHAPTER 2: LITERATURE REVIEW

2.1 The spine system

When simplified, the spine system can be considered an inverted pendulum (Reeves, Narendra, & Cholewicki, 2007), in that each joint must support and stabilize superior segments. Theoretically, when considering this type of system, any size of perturbation will cause the pendulum to permanently deviate from its state of equilibrium. If the goal of the system is to remain upright (the primary objective of the spine system), then this system will fail.

As the human spine does not buckle or collapse with every perturbation, there are obvious control mechanisms that are recruited to maintain a desired posture or movement. These control mechanisms are aspects of the active, passive and neural subsystems that operate together to achieve a motor outcome (Panjabi, 1992). In the simplest example, the spine's equilibrium is perturbed and this change in equilibrium would be detected by components of the neural subsystem (e.g. muscle spindles and Golgi tendon organs). Following detection, a movement response is created through a combination of reflexive properties and voluntary control to return the spine to a desired state. This movement response is communicated to the active subsystem (spine musculature) which then contracts to apply the necessary forces onto the passive subsystem (e.g.

tendons and bones). As a result of this control process, there is a movement outcome that is once again detected and modified per the same system, if necessary.

Perfect performance of this system would be achieved if all three subsystems operated flawlessly and all movements were achieved without excessive forces on tissues and subsequent pain. However, as with all human movement, the response of the system is never perfect and the speed and accuracy in which the system corrects changes in the perturbed state can be considered the system's performance. Due to the interaction of the subsystems, degeneration of one of the subsystems could lead to change and compensation in the others. This can cause error in movement, which has the potential to result in control outcomes that place tissues under excessive stress and cause injury (Cholewicki & McGill, 1996).

Considering the many degrees of freedom and dynamic demands of every-day spine tasks, it is clear that the spine system is very complex. As a result, it is extremely difficult for researchers to study in detail how control of the spine is achieved in healthy individuals and how control strategies differ when an individual is in pain. Another hurdle for researchers is deciding how to simultaneously observe and integrate all the subsystems to thoroughly understand spine movement and its control mechanisms. This has led to the development of different techniques to quantify control. Although there are advantages to having multiple methodologies to assess control, such as diversity of outcomes and details into different subsystems, there are disadvantages when comparing findings from different techniques. Without understanding the relationship between the outcomes of each technique researchers/clinicians will never know the degree of overlap between each tool in quantifying spine motor control. If these tools are to be used in a complimentary fashion, the overlap (or lack thereof) between each output will need to be understood before estimating spine control/function.

2.2 Impact of LBP

Globally, LBP causes the most years lived with disability when compared to 291 other conditions, making it the greatest contributor to global disability (Hoy et al., 2014). LBP is also a persistent condition, with recurrence rates in Canada being 20% within 1 year and reaching 36% over a 3 year period (Andersson, 1999). After ischemic heart disease, low back and neck pain were the second highest cause of disability-adjusted life years in high-income countries (Hurwitz, Randhawa, Yu, Côté, & Haldeman, 2018). Of additional concern, is the fact that LBP has shown a higher prevalence among older populations (Hurwitz et al., 2018). Considering the aging population globally, the prevalence of LBP will likely increase around the world.

Economically, the prevalence of LBP places a large burden on the health care system. The total cost of LBP encompasses both direct and indirect costs. Direct costs originate from things such as health care interventions and workers compensation claims, whereas indirect costs are costs associated with loss of employment and household productivity due to LBP. Medical expenditures alone for LBP are estimated to cost between $6 and $12 billion dollars annually in Canada (Bone and Joint Canada, 2014). These associated costs are in addition to loss of productivity and disability payments. Considering the prevalence and high costs associated with LBP globally, it is obvious that there needs to be an effort to understand this problem with hopes of developing solutions that will reduce the burden on the health care system.

2.3 LBP risk factors

LBP is understood to be multifactorial. The biopsychosocial model has been proposed to organize all the risk factors into three main categories: biological, psychological and social.

Within the psychosocial aspect of the model, many specific risk factors are identified. For instance, pain-related fear has been shown to be a factor in chronic LBP, as it can cause individuals to develop protective and restrictive movement behaviour by stiffening of the spine. This change can increase tissue loading due to co-contraction (van Dieën, Reeves, et al., 2018), which consequently can cause prolonged disability. Pain and disability can also lead to depression, which has been identified as a strong, independent predictor for an incident of LBP (Carroll, Cassidy, & Côté, 2004). Although the association between depression and LBP does not identify the causal relationship, it is speculated that depression may affect coping behaviour, which could be an intermediate step between depression and pain experience. In addition, factors such as feeling sad, exhausted and overwhelmed were strongly related to LBP prevalence among college students (Kennedy, Kassab, Gilkey, Linnel, & Morris, 2008). In this study, it was proposed that a potential mechanism to explain pain experience is through shared neurological pathways for the controlling of mood and perception of pain.

Despite the relationships between LBP and psychosocial characteristics, it is also important to consider biological risk factors of LBP. A major biological risk factor for LBP is the ability to control one's spine posture and/or movements. Spine motor control is achieved through complex interactions between the spine's active musculature, passive structures (bones, tendons and ligaments) and neural components (Panjabi, 1992). If there are impairments in any system, it could lead to undesired movements and the unsuccessful transmission of loads (Cholewicki & McGill, 1996). Specifically, it has been suggested that the sudden need to regain spine stability during a movement may cause excessive forces and tissue overload, resulting in injury (Cholewicki & McGill, 1996).

Stability of the spine is a term that has been loosely defined within biomechanical literature (Reeves et al., 2007). Simply put, a system is stable if it returns to an equilibrium state following a perturbation. Conversely, a system is unstable if it does not return to its equilibrium state after a disturbance. While the spine system is made stable through the interaction of subsystems, the performance of the spine system is measured by the speed and accuracy in which it can return to a state following a perturbation. This performance has been termed recently as "robustness" (Reeves et al., 2007) and can refer to the ability of the spine to return to a desired movement trajectory (dynamic) or postural state (static) following an internal or external perturbation. External perturbations describe disturbances from the external environment, whereas internal perturbations are disturbances that occur naturally during movement. As we experience internal and external perturbations during everyday movements, the ability to accurately recruit a movement control response may reduce the magnitude of the necessary reaction forces resulting from neuromuscular control error. Therefore, control of one's spine is critical to reducing forces on tissues and, as a result, reducing the risk of injury and pain (Reeves et al., 2007). This is especially important as the spinal column is inherently unstable. Evidence from cadaveric studies support this by demonstrating that the human osteoligamentous lumbar spine will buckle under compression loads of just 88 Newtons (Crisco, Panjabi, Yamamoto, & Oxland, 1992). As the passive loading from the upper body is much greater than this, the osteoligamentous structures alone are insufficient to keep the trunk upright. Therefore, there must be a contribution from the neuromuscular system that provides the robustness needed to maintain posture and control movement. If the combination of the osteoligamentous (passive) and neuromuscular (active and neural) subsystems are unable to provide adequate forces on the spine, movement can become impaired. An example of changes in the subsystems affecting movement control is presented in Howarth et al. (2013), where a viscoelastic change (creep) was induced on the passive tissues of the spine. After these changes were induced, there was a nonsignificant trend towards immediate loss of movement control followed by a return to baseline throughout a repetitive flexion protocol

(Howarth, Kingston, Brown, & Graham, 2013). It was suggested that this recovery of movement control was due to the neuromuscular system compensating for loss of stiffness in the passive tissues of the spine. Further, tissue creep has been shown to affect tension sensory thresholds of ligaments (Solomonow, 2004). As a result, alterations in the properties of the passive subsystem tissues can lead to changes in movement control and dysfunctional reflexive activity of muscles.

Peripheral muscular fatigue is another factor that can influence neuromuscular control of the spine through changes in the stiffness properties of muscles. Specifically, muscular fatigue has been shown to reduce active muscle stiffness, which results in a demand for larger reactionary force contribution to maintain necessary trunk stiffness (Granata & Slota, 2004). Specifically, this increased force contribution of paraspinal muscles has been shown to modify spinal stability and cause increased compressive forces on the spine through increased co-contraction of antagonist muscles. LDS has also been shown to decrease following fatigue of the torso extensor muscles (Granata & Gottipati, 2008). As LDS does not provide insight into the exact mechanism that causes this loss of local stability, it is impossible to tell where this loss of control is coming from. Granata & Gottipati (2008) suggest that it is possible that the reduction in control is due to the effect of fatigue on muscle recruitment patterns, neurological control, or both. Moreover, repetitive loading of the spine's passive tissues can increase their laxity, which can compromise control of the spine and increase injury risk when combined with decreased force production from active tissues (Solomonow, 2012).

It is clear that there are many factors that can affect the motor output of the spine. Without the ability to identify risk factors related to spine motor control, the question of how people are getting injured becomes a difficult issue to solve. Therefore, it is important that researchers have tools to not only quantify overall motor control performance, but also the contribution to control from different subsystems. This way, impairment within these systems can be recognized and addressed during rehabilitation to recover motor control and reduce the risk of injury.

2.4 LBP and motor control

Previously, there has been conflicting evidence regarding trunk motor control within LBP populations. It is known that there are alterations in control with occurrence of LBP; however, whether altered spine motor control is a cause or effect of LBP is debated. This debate is largely influenced by case-control and cohort studies being unable to determine causation of injury, as well as inconsistent findings within the literature. For instance, within the LBP population, postural sway was found to be increased compared to matched healthy controls (Radebold et al., 2001). Postural sway, in this case, was quantified using centre of pressure (COP) displacement during an unstable sitting test. On the other hand, no difference in magnitude of COP displacement was found between LBP and healthy populations during a similar seated balance test (van Dieën, Koppes, & Twisk, 2010). In this same study, smaller COP excursions were discovered in patients that had recently experienced LBP. Although both of these findings were during static tasks, similar conflicting evidence was found during dynamic movements. For example, Ross et al. (2017) conducted a study on healthy individuals with pain induced through nociceptive stimulation and found no overall differences in LDS during a repetitive spine flexion task between no pain, pain induced, and recovery conditions. Interestingly, differences were seen when pain catastrophizing (PC) was considered and individuals with high PC demonstrated reduced maximum Lyapunov exponents (i.e. tightened their control) whereas those with low PC showed increased maximum Lyapunov exponents (i.e. loosened control) following induced pain (Ross et al., 2017). Conversely, during a trunk repositioning task, LBP participants as a whole demonstrated modifications in movement time, peak movement velocity and acceleration (Descarreaux, Blouin,

& Teasdale, 2005). In addition, a group difference was found in participants with induced LBP, as they demonstrated decreased local stability during a repetitive spine flexion task (Ross, Mavor, Brown, & Graham, 2015). There could be many reasons why research is inconsistent within this area of research, such as subgroups within the LBP population (e.g. those that "tighten" control vs. those that "loosen" control (van Dieën, Reeves, et al., 2018)) or the studied parameters of motor control. As a result, summarizing and comparing findings is inherently difficult as there is no consistency in motor control quantification techniques. This brings forward the importance of utilizing methodology that classifies motor control similarly or understanding the relationship between the results of different techniques. An analogy to this issue, referenced by Reeves, Narendra and Cholewicki (2007), is to a poem by John Godfrey Saxe about six blind men and an elephant. In this poem, the blind men are given the task of identifying an object (i.e. the elephant) in front of them. Each man reaches out and touches a different part of the elephant; one touches the trunk while another touches the leg etc. In the end, all the blind men think that they are touching different things. For instance, the man touching the trunk perceives a snake while the man touching the leg perceives a tree. While no man is incorrect, as all of their decisions are supported by their perceptions, they are failing to observe the elephant as a whole (Reeves et al., 2007). Using this analogy, Reeves and colleagues explain how stability has the potential to be the elephant of spine biomechanics and outlines the importance of understanding what researchers are observing using different techniques so that they can have a broader view and observe the whole problem to understand what it truly is.

2.5 Local dynamic stability

LDS is one specific technique of assessing spine motor control, which utilizes cyclic kinematic data in order to quantify the spine's response to internal perturbations, which occur naturally during dynamic movements. This technique has been used frequently to identify spine control during repetitive lifting tasks (Graham, Sadler, & Stevenson, 2012) and repetitive unloaded trunk movement tasks under different conditions such as experimentally induced pain (Ross et al., 2015), athletes with LBP (Graham, Oikawa, & Ross, 2014) and muscular fatigue (Granata & Gottipati, 2008). In addition to spine flexion and extension, LDS was used to assess motor control differences in asymmetrical movement tasks (Graham et al., 2014) and during rotational and complex tasks that involved movement in the frontal and transverse planes (Dupeyron, Rispens, Demattei, & van Dieën, 2013). Both studies demonstrated increased local stability during movements outside of the sagittal plane (i.e. asymmetrical and complex movements), suggesting that movement control is task dependent. Therefore, when using LDS as an assessment technique, it is important to consider trunk movement in all movement planes as movement direction has been shown to influence dynamic control (Granata & England, 2006), and complex movements have been suggested to increase the risk of low back injury (Fathallah, Marras, & Parnianpour, 1998). As a result, current research groups are using trunk movements in all planes to assess dynamic trunk control in healthy and LBP individuals.

LDS is an analysis technique that can be used to evaluate chaotic behaviour within cyclic biological data. To quantify lumbar spine movement control, LDS uses the maximum finite-time Lyapunov exponent (λ_{max}), which is the maximum rate of divergence of a data point from an attractor/trajectory. An important step in LDS analysis, is the definition of a state space. A valid state space must contain state variables that unequivocally define the state of the system (i.e. the spine system) at any point in time (Dingwell, 2006). There are multiple accepted ways that the state space can be defined when using lumbar spine data. Specifically, the cyclic signal being used (angles, velocities, accelerations) can be changed, as well as the reconstruction dimension and utilisation of time-lagged signals. Previously, λ_{max} has been calculated from a 6-dimensional state space reconstructed using the Euclidean norm of the three lumbar spine angles (flexion-extension, lateral flexion and axial rotation) and its time-delayed versions (Beaudette et al., 2014; Graham & Brown, 2012; Graham et al., 2014; Granata & Gottipati, 2008). Another method to calculate λ_{max} during repetitive trunk movements, employed by Dupeyron et al. (2013), used a reconstruction dimension of 12 and reconstructed the state space using the raw linear and angular velocities of the trunk relative to the pelvis and their time delayed copies. Previous work comparing the results of time-delayed embedding and redundant state space definitions has demonstrated that, although trends exist, direct numerical comparisons between different state space reconstruction techniques should be made with caution (Gates & Dingwell, 2009). Additionally, the advantages that one reconstruction technique provides over another is unclear, and there is no universally accepted way to design the multidimensional state space (Dingwell, 2006). Therefore, it is important to consider different state space reconstruction techniques when using LDS, as multiple techniques are used within the literature and this is known to potentially influence λ_{max} estimations.

Regardless of state space definition, the subsequent steps in a LDS analysis remain the same. Within both state spaces, nearest neighbours are identified as two data points that are closest together on separate cycles, and the diverging/converging Euclidean distance between these points is tracked forward in time. The average rate of divergence of all nearest neighbour pairs across an entire cycle is then plotted, and a line of best fit is estimated from specific regions of this curve (i.e. from 0-0.5 cycles) (Bruijn, van Dieën, Meijer, & Beek, 2009a; Graham, Sadler, et al., 2012). To classify the short-term response to local perturbations, the average divergence over a complete cycle is considered and a line of best fit is attached to the first 50% of the divergence curve (Bruijn et al., 2009a). It is also important to consider that this portion of the divergence curve is generally linear to provide a stable estimate of the slope. The slope of this line is the resulting value of λ_{max} and is a representation of how a person (i.e. their spine system) returns to their original movement path following a local disturbance (Graham, 2012).

LDS is an advantageous movement control assessment technique as it allows control to be quantified during dynamic movements. Therefore, it can be applied in practical cyclical movements that occur in every-day life such as repetitive lifting or walking. Additionally, LDS analysis can be conducted using only a single repetitive data series (e.g. kinematics, kinetics, EMG, etc.). As a result, data collection techniques can include motion capture systems or small inertial measurement unit sensors. Assessing LDS of the spine has demonstrated strong reliability of measurement with a between-day intraclass correlation coefficient of 0.73 (Graham, Sheppard, Almosnino, & Stevenson, 2012). Previously, the versatility of data collection methods for a LDS analysis has been utilized to assess the risk of falling in patients with neurological disorders (Reynard, Vuadens, Deriaz, & Terrier, 2014). This study assessed LDS of patients in the clinic using an accelerometer to track accelerations of the trunk during a gait task. This diversity of equipment that can be used to collect data justifies the practicality of LDS for use in clinics or applied research, where complex software, equipment and expertise may not be readily available. Therefore, LDS can be applied in the clinic as a movement screening tool where patients can be initially assessed and treatments or further assessment plans can be individualized to improve the level of care (Beange, Beharriell, Wai, & Graham, 2017).

While LDS is able to consider non-linear data during dynamic movement tasks, it provides a “black box” assessment of the underlying control. Typically, when assessing LDS of the spine, the analysis only relies on kinematic data of the spine. This means that although LDS can identify tight (constrained movement) or loose (uncontrolled movement) control, there is less detail into what is causing the loss or recovery of it. Despite being unable to identify the underlying mechanisms with a Lyapunov analysis, the method still provides the advantage of accessibility as it does not require large amounts of equipment or set up time.

2.6 Systems identification

Within the literature, SI is another method that is used to assess neuromuscular control of the spine in an upright posture. Although SI methods were originally applied to the human body to assess muscle spindle feedback properties of the shoulder (Schouten, Vlugt, Hilten, & Helm, 2008), they were later applied to the spine as part of an effort to enhance motor control assessment in LBP patients. When SI techniques were employed to assess control in a LBP population, it was found that there was impaired reflexive adaptation with LBP and that the LBP population could be separated into subgroups to define different motor behaviour adaptations (van Drunen et al., 2015). In healthy populations, SI has been used to characterize the importance of reflexive contributions to upright trunk stabilization. Moorhouse & Granata (2007) used these techniques to demonstrate that reflexive activity was responsible for 42% of the total stabilizing trunk stiffness, reflecting that intrinsic muscle stiffness alone was unable to stabilize the spine. This demonstrates the importance of using SI to consider reflexive activity when assessing control behaviour of the spine. To assess control, SI applies known perturbations to the spine system and measures the response during the disturbances. Further, SI assessment can be used to quantify an individual's maximum or natural ability to control their spine depending on task instruction. To do this, participants complete the dynamic disturbance under two conditions where they are instructed to either resist the perturbations maximally (maximum control) or to relax their trunk during the perturbations but remain upright (natural control). With these data, the relationship between the input and output of the system is represented using frequency response functions (FRFs) and used to infer the contribution of the ligamentous, muscular and reflexive spine systems to stabilization of the low back (van Drunen, Maaswinkel, van der Helm, van Dieën, & Happee, 2013). SI uses closed loop assumptions (Schouten et al., 2008), which allows the use of FRFs, also known also as transfer functions, to provide detail into the frequency relationship between two signals.

SI uses a linear actuator to apply dynamic disturbances to the spine at the level of the 10th thoracic vertebrae (T_{10}) while the individual is in a seated and kneeling posture. This posture limits the movement of the pelvis and allows for the study of the trunk in isolation. Also, to not disrupt the closed loop assumption, the dynamic disturbance signal is pseudorandom to avoid muscular contractions due to predicting the perturbation (Maaswinkel, Griffioen, Perez, & van Dieën, 2016). As perturbations are applied at specific frequencies, the bandwidth of frequencies must be large enough to capture a wide dynamic range; although it is known that full power perturbations at high frequencies suppress an individual's reflexive response due to existing time-delays in reflexive pathways (Frans, Helm, Schouten, Vlugt, & Brouwn, 2002). Therefore, the reduced power method is applied in order to eliminate this interference and the power of the perturbations above 4 Hz is reduced to 40% (Mugge, Abbink, & Frans, 2007). During the perturbation period, contact force, displacement of the actuator and sEMG data are recorded, and the relationships between these variables can be represented as FRFs to determine the relationship between the input and output of the system. Specifically, admittance (kinematics) is described as the actuator displacement as a function of contact force, meanwhile reflexes describe the sEMG data as a function of the actuator displacement (Maaswinkel, van Drunen, Veeger, & van Dieën, 2015). Because the perturbation is applied over a bandwidth of frequencies, admittance and reflex responses are calculated at every frequency within the bandwidth. Responses at low frequencies (< 1 Hz) represents the response from intrinsic stiffness and reflexive properties, while the high frequency response (> 4 Hz) is modulated by mass of the trunk and contact dynamics. The response in the intermediate frequencies mainly represents intrinsic damping and reflexive responses. The reflex FRFs, derived from the sEMG signal and actuator displacement represents muscle spindle position, velocity, and acceleration feedback gains (van Drunen et al., 2013). This method of quantifying neuromuscular trunk control was shown to be a reliable method for the assessment of healthy individuals with good and fair test-retest intraclass correlation coefficients (ICCs) of 0.66 and 0.44 for admittance gain and reflexes, respectively. For LBP individuals, good test-retest ICCs of 0.73 and 0.67 were demonstrated for admittance gain and reflexes, respectively (Griffioen, Maaswinkel, Zuurmond, van Dieën, & Perez, 2016). SI has also been utilized within the literature to assess the effect of vision and posture on trunk neuromuscular control (Maaswinkel et al., 2015). This study concluded that visual feedback does not affect the neuromuscular response to small amplitude trunk perturbations.

SI is advantageous, as the admittance and reflexive FRFs can be used as inputs to a neuromuscular control model, which can quantify physiological properties that compose the intrinsic and reflexive contributions to stabilization of the low back (van Drunen et al., 2013). Intrinsic contributions describe lumbar stiffness and damping, whereas reflexive contributions consider lumbar muscle spindle position, velocity and acceleration feedback gains. The outputs from this model give insight into how they contribute to stabilization of the low-back and can identify where loss/compensation of control is coming from.

Another advantage to SI is that it provides extensive insight into the contribution of different sub-systems to postural control (van Drunen et al., 2013). For instance, by simultaneously quantifying both intrinsic and reflexive contributions, it provides a more accurate estimate of postural control than other methods that isolate one or the other. This allows for motor control deficits to be clearly identified that may not be detected by using other less detailed methods. In addition, as the FRFs can be used in a neuromuscular control model, SI can be used to quantify additional intrinsic and reflexive contributions to postural control such as the trunk's intrinsic stiffness, damping properties and muscle spindle position, velocity and acceleration feedback gains. This allows for more insight into how control is achieved and can be used to make a more accurate classification of trunk postural control.

Due to the linear assumption, it is not clear whether results from this type of analysis can be applied to more functional movements that involve motion in all planes. As the participant is placed in a seated kneeling posture, the results from SI convey the ability of the participant to remain in that static, upright posture. Although all movement has inherent neuromuscular control error and noise (Scott, 2004), this error may be different with changing task demands during dynamic movements. Therefore, it is important to understand if results from SI capture how an individual responds to noise and neuromuscular error that occurs during dynamic movements.

2.7 Summary

The association between LBP and motor control of the trunk is well established; however, the exact causal relationship is not well understood (van Dieën, Reeves, et al., 2018). The different methods that researchers use to quantify spine control may contribute to this uncertainty if these methods do not classify motor behaviour comparably. Therefore, the purpose of this thesis is to understand the relationship between motor control outputs of SI and LDS, two commonly used quantification techniques. It was hypothesized that SI control outcomes under relax task instructions will closely relate to LDS outcomes, as LDS also quantifies a natural movement response to perturbations. When considering all spine movement planes, it was also hypothesized that LDS during repetitive forward spine flexion will demonstrate the strongest relationship with SI outcomes, as the majority of movement is in the sagittal plane in both tasks.

CHAPTER 3: METHODS

3.1 Participants

Nine male and six female participants were recruited to participate in this study. Participants' mean age, height, mass and the associated standard deviations (SD) were 35.5 years (SD = 12.5), 175.4 cm (SD = 8.5) and 72.9 kg (SD = 11.6) respectively. Participant demographics (separated by sex) are presented in Table 1. All participants were healthy and had not experienced an episode of LBP within the past year, or a major musculoskeletal injury within the past six months before participation. In addition, all participants were free of any: cardiovascular conditions, neurologic disorders (neuropathy, neurodegenerative conditions), low back pain (discogenic, mechanical, myofascial), ankle injury (sprained, fractured) use of medication (anti-inflammatories, analgesics, anticonvulsants, antidepressants), history of low back injury (discogenic mechanical) use of anticoagulent therapy, stroke or TIA, spine trauma, motor vehicle accident, lumbar spine surgery, hypertension, CTD and focal neurological symptoms (sensory/motor). Participants were informed of this inclusion criteria before giving informed, voluntary consent.

Table 1. *Participant demographics.*

	Height (cm) Mean (SD)	Mass (kg) Mean (SD)	Age (years) Mean (SD)
Male	179.2 (7.0)	77.6 (10.0)	34.1 (12.0)
Female	169.6 (7.0)	65.7 (9.9)	37.5 (13.0)
All	175.4 (8.5)	72.9 (11.6)	35.5 (12.5)

All participants were recruited from the general university employee and student population. All participants met the inclusion criteria outlined in the participant information letter (Appendix A). SD = standard deviation.

3.2 Consent

Before any data were collected, this project was reviewed and approved by ethical review boards at both the University of Ottawa and Vrije Universiteit Amsterdam (University of Ottawa ethics #H02-17-11 and Vrije Universiteit ethics #VCWE-2017-146; Appendix A). Participants first read and signed a consent form (Appendix A) in the language of their choice (Dutch or English). To elaborate on procedures described in the form, a physical demonstration of the protocol was provided before the participant consented to participation.

3.3 Protocol

Following arrival to the Amsterdam Movement Sciences laboratory and completion of informed consent, demographic data (height, weight and age) were collected. If the participant was wearing loose fitting clothing, athletic-wear was provided for the duration of the data collection period. Once the participant was outfitted properly, they began completion of either task A (LDS) or task B (SI) in a randomized order.

3.3.1 Task A: local dynamic stability

For this task, participants were outfitted with two infrared active marker clusters, each equipped with three kinematic markers. One cluster was placed on the trunk over the T_{10}–T_{12} vertebrae (to track thorax movement), and the other on the first sacral vertebrae (to track pelvis movement; Figure 4A). The three-dimensional position of these segments were tracked using an Optotrack 3020 motion capture system at a collection frequency of 100 Hz to allow for computation of lumbar spine Euler angle data.

In addition to the marker clusters, the location of anatomical landmarks on the trunk and pelvis were marked using a digital pointer (Appendix B). This enabled the calculation of segment coordinate systems that were tracked by the position of the marker clusters.

Prior to beginning the protocol, the participant was centred in the motion capture area and instructed to stand comfortably with their feet shoulder-width apart. Following this, the position of their feet was marked so that they remained in the same position throughout the entire data collection period.

Task A protocol involved participants completing three blocks of repetitive trunk movement tasks in a randomized order (Dupeyron et al., 2013). The first block consisted of 35 cycles of repetitive forward flexion and extension (FE block; Figure 1). Specifically, participants touched a target at shoulder height and knee height repeatedly with a movement frequency of 15 cycles per minute (4 seconds/cycle or 0.25 Hz) to a metronome playing at 0.5 Hz (one beep at top and bottom), which is similar to the preferred movement frequency during the repetitive movement tasks presented in Dupeyron et al. (2013). During this block, participants were instructed to keep their arms outstretched directly in front of them with one hand on top of the other.

Figure 1. *Sequence of trunk movements to complete the FE movement block.*

For the second block, participants completed 35 cycles of repeated rotation (rotation block; Figure 2) alternating between touching a target placed on their right side with their left hand and a target placed on their left side with their right hand. Targets were placed at shoulder height and at an arm's length away. Movements were completed in synchrony with a metronome beat at 0.5 Hz. This replicated the movement frequency of the FE block (4 seconds/cycle or 0.25 Hz).

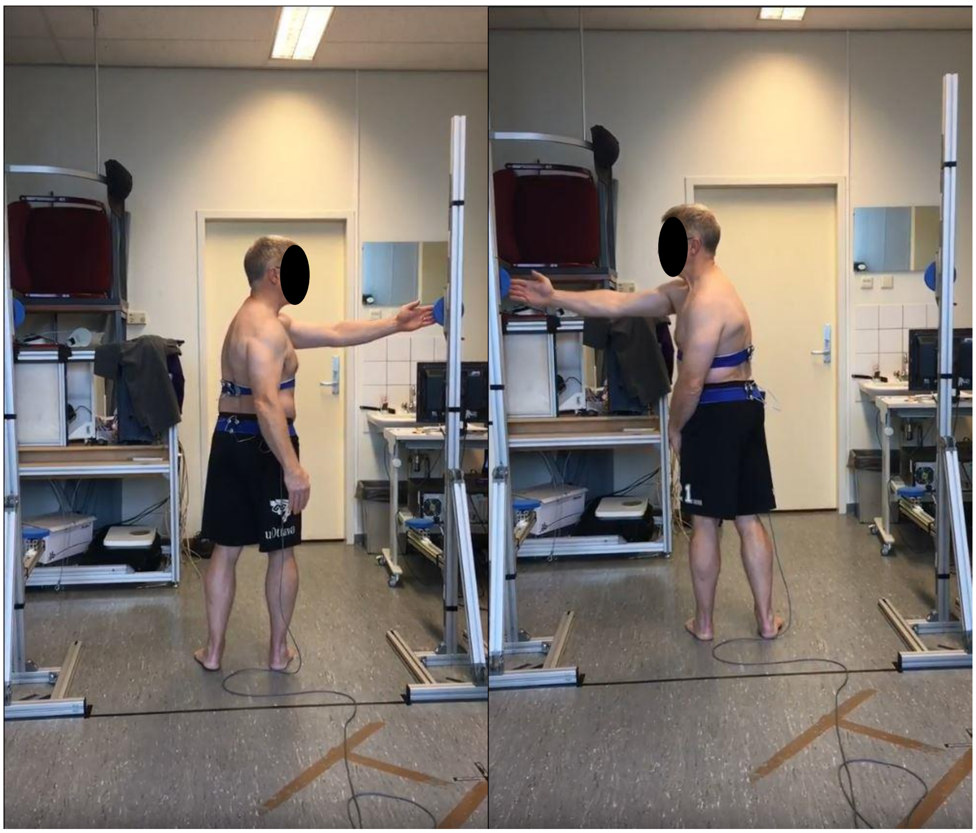

Figure 2. *Sequence of trunk movements to complete the rotation movement block.*

The third block (complex block; Figure 3) consisted of the participant completing 35 cycles of touching a target located at shoulder height on the right side, shoulder height on the left side, knee height on the right side and finally knee height on the left side consecutively to the beat of a metronome at 0.5 Hz. This resulted in a movement frequency of 8 seconds/cycle or 0.125 Hz. Although these movement tasks required minimal effort, rest periods in between blocks were provided to eliminate any effect of fatigue.

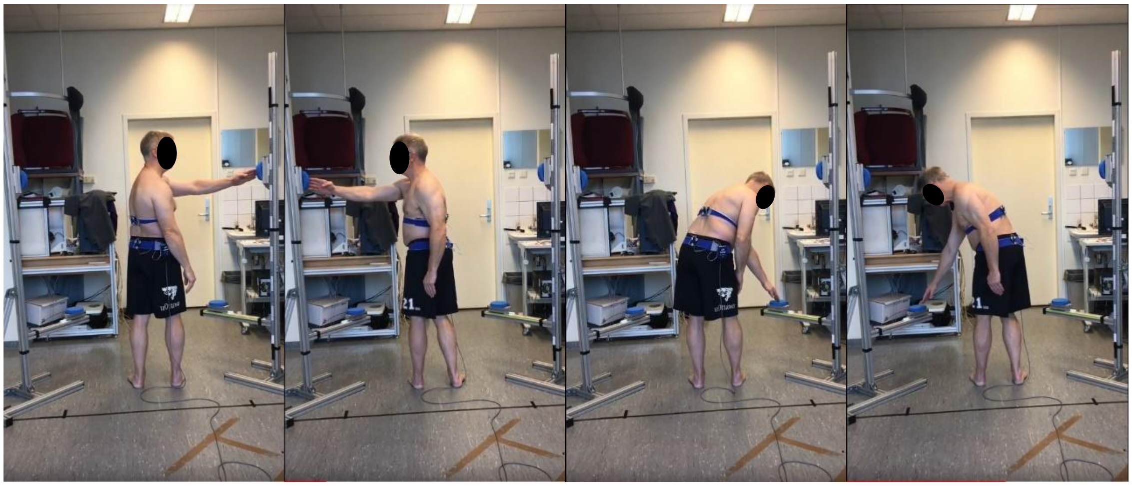

Figure 3. *Sequence of trunk movements to complete the complex movement block.*

3.3.2 Task B: systems identification

In this task, participants were first outfitted with sEMG sensors (sEMG REFA, TMSi, the Netherlands). Preparation of the skin involved shaving of any body hair and cleaning the skin with a cotton swab and alcohol. Pairs of sEMG electrodes (Ag/AgCl, inter-electrode distance 25mm) were placed on the longissimus (thoracic and lumbar) and iliocostalis (thoracic and lumbar) muscles according to Willigenburg, Kingma, & Van Dieën (2010). Specifically, the electrodes were placed: 3 cm lateral to the midpoint of the L_4 and L_5 vertebrae (lumbar longissimus), 6 cm lateral to the L_2 vertebra (lumbar iliocostalis), 6 cm lateral to the T_{11} vertebra (thoracic iliocostalis) and 4 cm lateral to the T_9 vertebra (thoracic longissimus; Figure 4B).

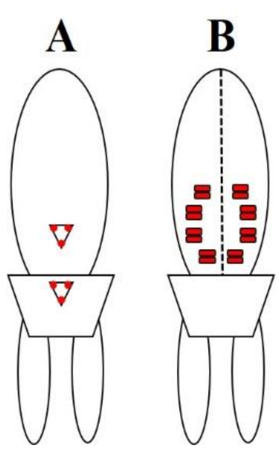

Figure 4. *Posterior view of torso and lower body. A) Location of active marker cluster for LDS task. B) Location of EMG sensors during SI task.*

These muscles were chosen as they have previously demonstrated high coherence with trunk movements (van Drunen et al., 2013).

Following attachment of sEMG sensors, participants were placed in a seated-kneeling posture, and a pelvis restraint was secured at the level of their anterior superior iliac spine to prevent any translation or movement of the pelvis. Participants were then blindfolded to eliminate any visual feedback to trunk proprioception and instructed to cross their arms across their chest for the duration of the trial (Figure 5). Once the participant was set up for the trial, a magnetically-driven linear actuator (Servotube STB2510S Forcer and Thrustrod TRB25-1380, Copley Controls, USA) was positioned level with T_{10} and in-contact with their spine. Following verbal confirmation that they were ready; the perturbation protocol began. Participants completed three, fifty-second-long dynamic disturbances in which the linear actuator applied pseudorandom force perturbations, in two different conditions (resist and relax). Each participant completed both the resist and relax task in a randomized order. Three trials in each condition were completed and values were averaged across trials to reduce noise and improve accuracy of measurement. Although the trials required minimal effort, a rest period was given between trials to ensure there was no residual effect of fatigue.

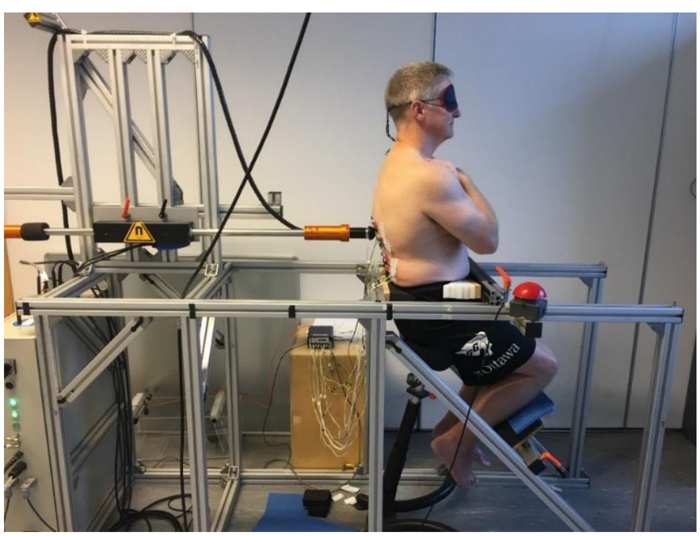

Figure 5. *Participant set-up for completion of Task B (SI).*

The pseudorandom force signal was composed of a ±35 N dynamic disturbance in addition to a 60 N preload for the actuator to maintain contact with the participant (as they are not attached). During the time to preload (3s) the participant was instructed to maintain a neutral, upright posture. In addition, the signal was composed using a crested multi-sine signal for a total of 20 seconds with 18 logarithmically distributed frequency pairs within a bandwidth of 0.2-15 Hz (van Drunen et al., 2013). To eliminate modified reflexive activity at high frequencies, the Reduced Power Method was applied and the power above 4 Hz was reduced by 40% (Mugge et al., 2007). In total, one trial lasted 50 seconds in duration; this included a ramp to pre-load (3s), a hold at pre-load (2s), a start-up period (the last 5s of the dynamic disturbance) to reduce transient behaviour, and two dynamic disturbances (20s each).

The two conditions in this task involved participants completing the dynamic disturbance protocol under resist and relax task instructions. Specifically, the resist instructions were to maximally resist the force perturbations and remain upright for the entire trial. This task instruction was given to measure the individual's maximum level of control. In contrast, the relax instructions were to relax the trunk completely; however, to remain upright following a perturbation. This task instruction was given to quantify a more natural response to mechanical perturbations, although it was important that the participant did not relax completely so that the perturbations did not push them over or cause too much rotation that the linear assumption would be violated. All outcome variables of SI were calculated under both the resist and relax task instructions for comparison to LDS outcomes.

van Drunen et al. (2013) found previously when using this perturbation apparatus that rotations occurred at both the spine and the pelvis, as the pelvic restraint was unable to completely immobilize the pelvis and isolate the trunk. Nevertheless, this movement in the pelvis was minimal and participants still demonstrated substantial trunk movement. In the same study, a kinematic analysis while using this apparatus also discovered that there was no well-defined rotation point necessary to define trunk rotations between participants. Therefore, kinematic responses to these perturbations were defined as translations by recording the displacement of the actuator probe throughout the trial (van Drunen et al., 2013). Contact force between the actuator probe and participant was also collected. Both displacement and contact force of the probe were collected at a frequency of 2000 Hz using an instrumented probe (Servotube position sensor & Force sensor FS6-500, AMTI, USA).

Admittance and reflexive FRFs were also used as inputs into a validated neuromuscular control (NMC) model (van Drunen et al., 2013). This NMC model was used to further quantify properties of low back stabilization, such as the lumbar intrinsic stiffness and damping and muscle spindle position, velocity and acceleration feedback gains. Specifically, the intrinsic contributions are grouped into lumbar stiffness and damping, which encompasses properties of muscle viscoelasticity and passive tissues within the lumbar spine. Feedback properties consist of proprioceptive feedback that is received from muscle spindles with regard to the position, velocity and acceleration of muscles.

3.4 Data analysis

3.4.1 Local dynamic stability

All kinematic operations were calculated using custom MATLAB software (R2017A, The MathWorks Inc., USA). Right hand local coordinate systems for the trunk and pelvis were first calculated based on the locations of anatomical landmarks identified using a digital pointer, and the three-dimensional location of these segments were tracked using their associated marker clusters. Three-dimensional lumbar spine angles were calculated using Euler rotation matrices (flexion-extension/lateral bending/axial twist sequence) of the trunk coordinate system relative to the coordinate system of the pelvis. Linear velocities of the trunk were defined as the first derivative of the trunk position relative to the pelvis position. Angular velocities were defined as the first derivative of the trunk Euler angle data. All kinematic data were first filtered using a 2nd order lowpass zero-lag Butterworth filter with a 10 Hz cut-off frequency. This cut-off frequency was chosen as it is well above the natural frequency of trunk movements (Moorhouse & Granata, 2005) and any signal higher than this frequency was considered noise.

Kinematic data from the FE and complex block were divided into cycles using the peak flexion-extension angles of the lumbar spine. In the rotation task, peak rotation angles were selected to divide the data into cycles. Of the 35 total cycles in each task, only the last 30 were selected for analysis in order to allow the participant to reach a steady-state movement pattern. Each block was normalized to a length determined by equation 1.

$$\#\ of\ samples = \#\ of\ cycles\ \times cycle\ time(s) \times collection\ frequency \quad (1)$$

Therefore, the FE and rotation blocks were normalized to 12000 data points, whereas the complex block was normalized to 24000 data points, as it is understood that the number of samples can have an effect on the LDS analysis (Bruijn, van Dieën, Meijer, & Beek, 2009b). It is important to note that by normalizing the entire length of the sample, cycle to cycle temporal variability is maintained.

As there are different methods of state-space reconstruction proposed within the literature, two main techniques were explored (6D and 12D). For the 6D technique, lumbar spine angles were first biased into a positive Cartesian space to remove any zero crossings and relative bias between movement planes (Beaudette, Howarth, Graham, & Brown, 2016). The state was defined similar to Graham et al. (2014) where a 6-dimensional state-space ($\mathrm{Y(t)}$) was reconstructed using the

Euclidean norm of the 3-dimensional lumbar spine angle (r) at each time point (t) and its time-lagged (T_L) versions as per Equation 2 (6D technique).

$$Y(t) = [r(t), r(t + T_L), r(t + 2T_L), \cdots, r(t + 5T_L)] \qquad (2)$$

For the 12D technique, the state was reconstructed similar to Dupeyron et al. (2013), where a 12-dimensional state-space ($Y(t)$) was reconstructed using the trunk's linear ($\dot{x}$, $\dot{y}$, $\dot{z}$) and angular ($\dot{\theta}$, $\dot{\varphi}$, $\dot{\psi}$) velocities and their time-lagged (L) copies as per Equation 3 (12D technique).

$$Y(t) = \begin{bmatrix} \dot{x}_1 & \dot{y}_1 & \dot{z}_1 & \dot{\theta}_1 & \dot{\varphi}_1 & \dot{\psi}_1 & \dot{x}_{L1} & \dot{y}_{L1} & \dot{z}_{L1} & \dot{\theta}_{L1} & \dot{\varphi}_{L1} & \dot{\psi}_{L1} \\ \dot{x}_2 & \dot{y}_2 & \dot{z}_2 & \dot{\theta}_2 & \dot{\varphi}_2 & \dot{\psi}_2 & \dot{x}_{L2} & \dot{y}_{L2} & \dot{z}_{L2} & \dot{\theta}_{L2} & \dot{\varphi}_{L2} & \dot{\psi}_{L2} \\ \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots \\ \dot{x}_n & \dot{y}_n & \dot{z}_n & \dot{\theta}_n & \dot{\varphi}_n & \dot{\psi}_n & \dot{x}_{Ln} & \dot{y}_{Ln} & \dot{z}_{Ln} & \dot{\theta}_{Ln} & \dot{\varphi}_{Ln} & \dot{\psi}_{Ln} \end{bmatrix} \qquad (3)$$

In both methods of state-space reconstruction, a time-lag of 10% of the cycle length was selected to match previous literature (Bruijn et al., 2009b; England & Granata, 2007; Granata & England, 2006; Howarth et al., 2013). Within both state-spaces, an algorithm developed by Rosenstein et al. (1993) was used to identify nearest neighbour trajectories and assess their exponential rate of divergence over time. The average rate of divergence between all nearest neighbours pairs was plotted over a period from 0-1 cycle, and the maximum finite-time Lyapunov exponent (λ_{max}) was represented as the slope of a line of best fit spanning 0-0.5 cycles (Graham, Sadler, et al., 2012). λ_{max} from all repetitive movement blocks and different state space reconstruction techniques were compared to the outputs from SI analysis to understand the relationships between them.

3.4.2 Systems identification

sEMG data collected during task B (SI) were zero phase, first order, high-pass filtered at 250 Hz before being rectified. This filter was chosen as it has demonstrated improved joint moment estimation in biomechanical modelling of the lumbo-sacral joint (Staudenmann, Potvin, Kingma, Stegeman, & van Dieën, 2007). In addition, although removing most of the raw signal, high pass filtering has been shown to improve muscle force estimation of the biceps brachii (Potvin & Brown, 2004).

All SI procedures were calculated using custom MATLAB software (R2017A, The MathWorks Inc., USA). Closed-loop identification techniques (Schouten et al., 2008) were used to represent the admittance ($H_{adm}(f)$; equation 4) and reflexes ($H_{emg}(f)$; equation 5) as FRFs. More specifically, admittance describes the actuator displacement ($X_A(t)$) as a function of the contact force ($F_c(t)$), while reflexes describe the sEMG ($e(t)$) data as a function of the actuator displacement.

$$H_{adm}(f) = \frac{S_{F_p X_A}(f)}{S_{F_p F_c}(f)} \qquad (4)$$

$$H_{emg}(f) = \frac{S_{F_p e}(f)}{S_{F_p X_A}(f)} \qquad (5)$$

Where $S_{F_p X_A}(f)$, $S_{F_p F_c}(f)$ and $S_{F_p e}(f)$ represent the estimated cross-spectral densities between the Fourier transformed force-perturbation ($F_p(f)$) and actuator displacement ($X_A(f)$), contact force ($F_c(f)$) and sEMG ($e(f)$), respectively (Maaswinkel et al., 2015). Cross-spectral densities were only calculated at the frequencies in which there was power in the perturbation signal. Densities were averaged across six time segments (three trials with two 20s dynamic disturbances) and over two adjacent frequency points. Lastly, $S_{F_p e}(f)$ was averaged over the left and right muscles.

Coherence of admittance (γ^2_{adm}) and reflexes (γ^2_{emg}) were calculated using equations 6 and 7, respectively (van Drunen et al., 2013).

$$\gamma^2_{adm}(f) = \frac{\left|S_{F_p X_a}(f)\right|^2}{S_{F_p F_p}(f) S_{X_A X_A}(f)} \qquad (6)$$

$$\gamma_{emg}^2(f) = \frac{\left|S_{F_pe}(f)\right|^2}{S_{F_pF_p}(f)S_{ee}(f)} \quad (7)$$

Coherence represents the relationship between the input and the output of the system. A coherence of 1 would represent a perfect relation between the input and output whereas a coherence of 0 would represent no relation between the two. Due to spectral densities being averaged over 12 points (2 adjacent frequencies x 6 repetitions), coherence of over 0.24 was considered significant at $p < 0.05$ (Halliday et al., 1996). Only coherence at the frequencies that reached this significance level were used for analysis. Admittance gains at five frequencies (0.23, 0.33, 0.48, 0.73 and 1.08 Hz) were used for comparison with LDS, as behaviour with the low frequency band (< 1.1 Hz) is modulated by task instructions. Behaviour outside of these frequencies is typically the same in the resist and relax task and therefore was omitted from further analyses.

Admittance and reflex FRFs were put into a NMC model to quantify additional intrinsic and reflexive properties of the spine (van Drunen et al., 2013). Specifically, the model quantified lumbar intrinsic stiffness (K) and damping (B), which consider the viscoelastic properties of muscles and passive tissues, as well as feedback gain from muscle spindle position (K_p), velocity (K_v) and acceleration (K_a). These parameters were calculated in both the resist and relax conditions.

3.5 Statistical analysis

Differences in λ_{max} estimations between the FE, rotation and complex blocks were assessed using a one-way repeated measures ANOVA considering task as a within-subject factor. This comparison was made for both 6D and 12D reconstruction techniques separately, in order to observe trends between movement blocks within each technique. Differences between blocks were considered significant if the p-value was less than the alpha cut-off of 0.05 ($p < 0.05$). If an overall statistically significant difference between movement blocks was established, a Bonferroni post-hoc test was used to determine the specific location of the differences.

A step-wise linear regression was used to define regression models between λ_{max} from each task and SI outputs. A step-wise regression is similar in procedure to the forward regression, where variables must meet an entrance criterion to be considered in the model. In contrast to the forward selection protocol, once a variable is included in a step-wise model, all other included variables are re-assessed and excluded from the model if the significance has been reduced below a specified tolerance level. Using the step-wise procedure, λ_{max} was assigned as the dependent variable (DV), and selected outputs from SI were included as independent variables (IV). As there are no limitations to the number of independent variables that can be used in a stepwise linear regression, only the variables from SI that were thought to be closely related to movement behaviour were chosen as potential predictors (van Dieën, van Drunen, & Happee, 2018). Therefore, the independent variables assigned were the admittance gain at 0.22, 0.33, 0.48, 0.73 and 1.08 Hz, muscle spindle (MS) position (K_p), velocity (K_v) and acceleration (K_a) feedback gains as well as lumbar intrinsic stiffness (K) and damping (B; Table 1). As per the stepwise regression procedure, any independent variable that fell below the probability of F (p-value) of $p < 0.2$ was considered for inclusion in the model. Once an independent variable was included in the model, if the probability of F (p-value) of any other previously included variables exceeded the cut-off of $p > 0.3$, that variable was removed from the model. Models were considered valid if the F-test significance was less than the alpha cut-off of 0.05 ($p < 0.05$). The adjusted R-square value for each model is presented as a percentage of the variance in λ_{max} that is described by the independent variables included in the model. As it was unknown which task instructions would best describe movement control during dynamic movements, λ_{max} was compared to SI outputs in the relax and resist tasks separately. Collinearity was assessed, and if the variance inflation factor (VIF) exceeded a value of 5, the model was not considered valid as it is likely that two independent variables considered in the model were strongly correlated.

Table 2. *Format for set-up of stepwise linear regression.*

DV	IV									
	Admittance gain (Hz)									
λ_{max}	0.23	0.33	0.48	0.73	1.08	K	B	K_p	K_v	K_a

CHAPTER 4: RESULTS

4.1 LDS

When using the 6D technique for estimating λ_{max}, there was a significant overall main effect of trial ($p < 0.001$) and post-hoc testing revealed that the FE movement block demonstrated the least locally stable behaviour when compared to the rotation (Rot) ($p < 0.001$) and complex (Comp) movement blocks ($p < 0.001$). There was no significant difference in λ_{max} estimations between the rotation and complex movement blocks ($p = 0.16$; Figure 6).

When using the 12D technique for estimating λ_{max}, there was an overall main effect for trial ($p < 0.001$) and post-hoc testing revealed that the FE movement block demonstrated the least locally stable behaviour when compared to the rotation ($p < 0.001$) and complex movement blocks ($p < 0.001$). In contrast to the 6D method of estimating λ_{max}, participants demonstrated greater local stability in the complex task than the rotation task ($p < 0.001$; Figure 6).

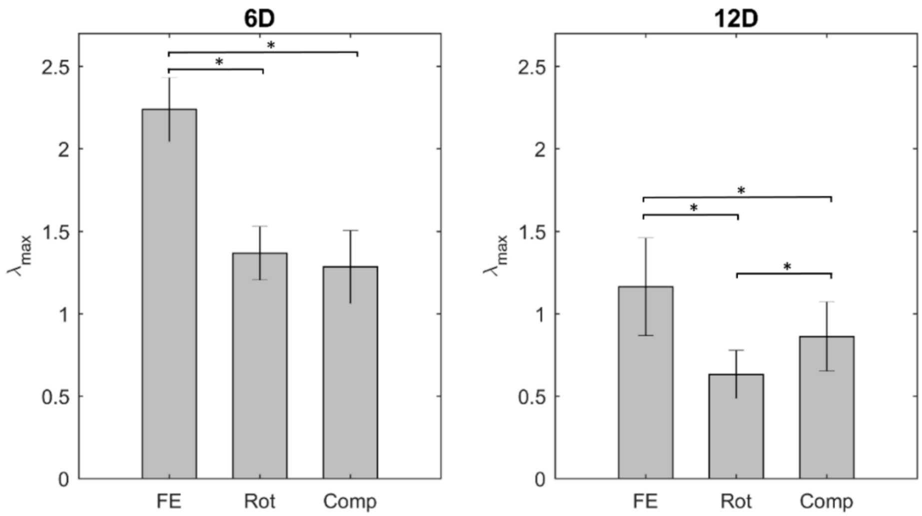

Figure 6. *λ_{max} estimations in both state space reconstruction techniques. Error bars represent the standard deviation about the mean. Black bars with an asterisk represent significant differences between conditions (p < 0.05).*

4.2 Regression models: 6D technique

Results presented in this section are linear regression models associated with λ_{max} calculations using the 6-dimensional state space composed of the Euclidean norm of the lumbar spine angle and its time-delayed copies. The strongest predictive models in each comparison are summarized in Table 3.

4.2.1 FE block

Admittance gain at 1.08 Hz (β = 355.1, p = 0.05) in the relax task predicted 20.7% (R^2 = 0.264, p = 0.05) of the variance in λ_{max} during the FE block.

In the resist task, all of the models violated the collinearity assumption (VIF > 5) or failed to reach the critical significance level and were considered insignificant.

4.2.2 Rotation block

In the relax task, the model that included MS position (β = 0.0001, p < 0.001) and velocity (β = 0.0003, p = 0.001) feedback gains and admittance gain at 0.22 Hz (β = 526.9, p = 0.003) and 0.73 Hz (β = -591.4, p = 0.007) was able to predict 64.7% (R^2 = 0.748, p = 0.005) of the variance in λ_{max}. All other models in this comparison violated the collinearity assumption (VIF > 5), or did not reach the critical significance level and were considered insignificant.

No SI variables in the resist task were able to significantly predict λ_{max} in the rotation block.

4.2.3 Complex block

No SI variables in the relax or resist task met the inclusion criteria to be considered as a predictor for λ_{max} values calculated during the complex movement block.

Table 3. *Strongest predictive models when using the 6D technique for λ_{max} estimation.*

		SI	
		Relax	Resist
	Flexion/Extension	20.7%[a]	No significant model
λ_{max}	Rotation	64.7%[b]	No significant model
	Complex	No significant model	No significant model

a. Predictors: Admittance gain at 1.08 Hz, (p = 0.050)
b. Predictors: Admittance gain at 0.22 and 0.73 Hz, K_p and K_v, (p = 0.005)

4.3 Regression models: 12D technique

All results presented in this section are linear regression models associated with λ_{max} calculations using a 12-dimensional state space reconstructed using the linear and angular velocities of the trunk relative to the pelvis and their time-delayed versions. The strongest predictive models in each comparison are summarized in Table 4.

4.3.1 FE block

No SI variables in the relax or resist task met the inclusion criteria to be considered as a predictor for λ_{max} calculations during the FE block.

4.3.2 Rotation block

No regression models in the relax or resist task met the critical significance level to be considered as a model to predict λ_{max} calculations during the rotation block.

4.3.3 Complex block

In the relax task, the model including admittance gain at 0.73 Hz (β = -842.8, p = 0.007) as well as lumbar damping (β = 0.0005, p = 0.009) was able to describe 42.0% (R^2 = 0.503, p = 0.015) of the variance in λ_{max}.

In the resist task, no models reached the critical significance level ($p < 0.05$) and therefore no models were considered for prediction of λ_{max}.

Table 4*. Strongest predictive models when using the 12D technique for λ_{max} estimation.*

		SI	
		Relax	Resist
	Flexion/Extension	No significant Model	No significant Model
λ_{max}	Rotation	No significant Model	No significant Model
	Complex	42.0%[a]	No significant Model

a. Predictors: Admittance gain at 0.73 Hz and B, (p = 0.015)

CHAPTER 5: DISCUSSION

The purpose of this thesis was to compare the motor behaviour outcomes of two popular spine control quantification techniques, namely, LDS and SI. Understanding the relationship between these outcomes can address inherent advantages and disadvantages of either technique and improve spine control assessment.

To broaden the findings of this exploration, λ_{max} calculations were completed using two different state-space reconstruction techniques. Although state spaces reconstructed using redundant information (i.e. derivatives) have been shown to produce larger error than methods using delay embedding, there is no universally accepted way to design the state space (Dingwell, 2006; Gates & Dingwell, 2009). Therefore, it is important to understand which method may be more closely related to SI outcomes. Specifically, the 6D method uses a 6-dimensional state space constructed using the Euclidean norm of the lumbar spine angle and its time-delayed copies (Graham et al., 2014; Granata & Gottipati, 2008; Ross et al., 2015). Additionally, λ_{max} was calculated using a 12-dimensional state space consisting of the three-dimensional linear and angular velocities and their time-delayed versions (Dupeyron et al., 2013).

As hypothesized, SI control outcomes in the relax condition demonstrated the strongest relationship with λ_{max} values. Using the 6D method to calculate λ_{max}, SI control outcomes in the relax task described 20.7% of the variance in the FE block and 64.7% in the rotation block. When using the 12D method of calculating λ_{max}, SI outcomes in the relax condition described 42.0% of the variance in λ_{max} during the complex block. Thus, all of the predictive models related to outcomes of SI in the relax condition. On the other hand, λ_{max} values (6D and 12D) in all movement blocks showed no relationship with SI outcomes in the resist condition. These results may be attributable to the nature of the task and the task instructions. For instance, the SI relax condition is designed to quantify a natural response to external perturbations, whereas the resist task is meant to quantify an individual`s maximum response to perturbations. Although participants were slightly constrained by the metronome frequency in the LDS task, there were no instructions given to maximize movement control. As a result, a more natural movement pattern is demonstrated. Therefore, both SI during relax instructions and LDS could be quantifying a natural response to kinematic disturbances.

In rejection of the second hypothesis, λ_{max} in the FE block was not as strongly predicted by the outcomes of SI as was λ_{max} in the rotation block. As the force perturbations during the SI task were in the anterior direction, similar control strategies were expected to be recruited during the FE block as the primary movement is in the anterior/posterior direction. This disagreement may be the result of the magnitude of the movements. In the SI task, the perturbations experienced by the actuator were very small in magnitude (25N minimum – 95N maximum) in order to maintain linear assumptions. Therefore, the participant mostly maintained an upright posture during the entire SI task, whereas the movement of the spine in the sagittal plane was very large during the LDS task. Previous investigation has shown that there are changes in muscle recruitment patterns throughout a trunk flexion and extension cycle (Leinonen, Kankaanpää, Airaksinen, & Hänninen, 2000). This is due to changing spine moments cause by a forward leaning posture, which requires different muscle contributions to oppose the force of gravity. Therefore, the difference in magnitude of the movements may have caused recruitment of different control strategies to accommodate modulated neuromuscular demands. In addition to different muscular demands necessary to complete each task, altered proprioceptive feedback may explain the differences in control outcomes. As predicted by the optimal feedback control theory, sensory feedback can be modified depending on task context (Scott, 2004). This suggests that feedback received during SI, where the task was maintaining a constant trunk position, is different than when the participant was moving during the LDS task. In this instance, the sensory feedback is provided by receptors within tissues. However, it is also important to consider the changes that may be associated with visual feedback. To maintain closed loop assumptions when using SI techniques, participants were blindfolded to eliminate the interference of visual feedback to the location of the trunk. As participants were not blindfolded during the LDS task, there would have been a visual feedback contribution as to the trajectory of the participant's trunk. Despite vision having a negligible effect on trunk stabilization with small amplitude perturbations, it is known that vision can contribute to compensatory movement through feedback information (Maaswinkel et al., 2015). It is possible that larger amplitude movements and vision in the LDS tasks altered proprioceptive feedback, thereby recruiting different motor control strategies when compared to the SI task.

During the rotation block, λ_{max} calculations using the 6D method demonstrated the strongest relationship with SI outcomes in the relax condition as the predictive model described 64.7% of the variance in λ_{max}. The explanation of the strong relationship between SI outcomes and λ_{max} estimations in the rotation block may be related to the similar trunk postures maintained in each task. In the SI task, the seated kneeling posture isolates the spine in an upright posture, which the participant maintains throughout the entire trial. In the rotation block during the LDS task, the objective is to touch targets on the right and left side of the body at shoulder height and an arms-length away. The goal of this task can be achieved using only axial rotation of the spine, and any flexion/extension excursions were therefore very small. As the passive stiffness of the trunk is known to increase with lateral and forward spine flexion (McGill, Seguin, & Bennett, 1994), the lack of motion in these directions during the rotation block may indicate that there were similarities with the SI task in the passive stiffness of the trunk. These comparable stiffness properties may have caused feedback information from receptors in the spine's tissues in the rotation block to resemble the information received in the SI task. In addition, similar upright postures would have provided like tactile feedback from mechanoreceptors in the skin. This cutaneous information can affect postural control, as it can be used to supplement other proprioceptive systems to determine the location and orientation of the trunk (Maaswinkel, Veeger, & van Dieën, 2014; Martin, Lee, & Sienko, 2015). It is possible, then, that the proprioceptive feedback in the rotation block was most similar to feedback in the SI task, and therefore elicited a similar neuromuscular control response. This equivalent response is reflected in the strong relationship between outputs.

LDS calculations using the 12D technique during the complex movement block demonstrated the next strongest relationship with SI outcomes. In this comparison, it was able to predict 42.0% of the variance in λ_{max}. The strength of this relationship may also be related to the nature of the postural demands during this task. Mainly, this movement included shared aspects of the rotation block when the participant touched a button at shoulder height on one side and the subsequent movement was to touch a button at shoulder height on the opposite side of their body. As explained previously, this movement may have produced a similar feedback signature leading to a selection of similar motor control strategies. However, as the relationship is weaker than the rotation block, it is important to consider the task differences. In addition to axial rotation, the complex block involved flexion extension movements as well as lateral bending. In flexed postures, the spine system demonstrates modulation of control behaviour (Maaswinkel et al., 2015). Specifically, when the spine is in the flexed posture, there is a decrease in admittance and reflexive contribution to postural control. As it is understood that reflexive behaviour is essential to providing adequate stiffness (Moorhouse & Granata, 2007), it is possible that changing contributions during spine flexion in the complex block resulted in different motor control strategy recruitment. As SI outcomes were able to predict higher percentages of the variance in λ_{max} in both the rotation and complex blocks and these movement blocks have shared postural demands, future research should assess the effect of maintained trunk posture during repetitive movement tasks. It may be that different postures elicit more/less of a response from different subsystems and it would be interesting to see if this response is reflected in the predictors from SI assessment. With this understanding, repetitive movement tasks could be designed to target specific subsystems and assess the ability of each to control movement.

The predictive models presented in this thesis can contribute to the overall understanding of spine control quantification and the interpretation of these assessments. Mainly, it can be said that there are shared spine movement strategies that are captured and quantified by both SI and LDS under very specific conditions. It is clear that these strategies quantify similar movement behaviour when LDS is estimated using a 6D reconstructed state space during a rotation task and SI outcomes are interpreted during relax task instructions. Therefore, if poor movement control is demonstrated during a repetitive rotation task, it is most likely that this level of control will be reflected in SI outcomes under relax task instructions. This concept is important to understand to improve clinical assessment of spine motor control. Although LDS provides no detail regarding the source of control, the data required can be collected using small, inexpensive equipment and is therefore considered applicable for clinical assessment. Therefore, LDS can be used initially in the clinic as a screening procedure and if risky movement behaviour is detected, SI assessment can be prescribed as a follow-up to clinical assessment to describe where detrimental movement control may be coming from. If however, no risky movement control is indicated by LDS assessment, attention can be given to physiological or psychosocial risk factors that may explain a current episode or risk of LBP. This system of assessment can help reduce unnecessary assessment strategies and individualize LBP treatment plans (Beange et al., 2017). It is important to consider that the use of only healthy individuals in this thesis does not allow for the identification of LDS thresholds that may identify “risky” movement behavior that would serve as cut-off points for further assessment. Previous findings suggest that there may be two phenotypes within the LBP population; patients that tighten (stabilize) their control to potentially avoid large movements and tissue strain, and patients that loosen (destabilize) control to potentially reduce forces on the spine associated with co-contraction and stiffening behaviour (Ross et al., 2017; van Dieën, Reeves, et al., 2018). Therefore, in the future, LDS should be measured using consistent methods on a number of participants with known motor control deficits to define these thresholds that might constitute risky behaviour. This is an important progression that needs to be addressed before this movement control assessment strategy can be effectively implemented within clinics.

In addition to consideration of strong predictive models, it is also important to interpret the lack of predictive models when comparing some outputs of SI and LDS. Although both methods quantify properties of motor control, in most conditions, movement control strategies are different, and these quantification techniques capture different behaviour. It is important to understand that motor behaviour is constantly influenced by noise (in both the ascending sensory and descending efferent information), task context and the nature of the sensory feedback itself (Scott, 2004). If there are changes to these parameters, it could lead to altered motor control output. When considering that SI techniques measure the response of the spine system in a relatively static, upright posture, whereas LDS assesses control during dynamic movements, it may be that parameters of noise, context and feedback are different. This would lead to the utilization of varying motor behaviour strategies to complete the demands of each task. Therefore, making direct comparisons from the outcomes of these techniques would be inaccurate and may lead to confusion when trying to identify specific motor control changes with conditions like LBP, especially when the presence of pain may directly affect proprioceptive information (Tong et al., 2017). This brings forward a methodological barrier to consistently quantifying motor control and outlines the need for a defined framework of motor control assessment. The results from this thesis can be used as a foundation to focus future exploration into the similarities between LDS and SI in different populations.

Within the overall technique of identifying LDS, there are multiple accepted ways of completing this analysis. Differences exist in the algorithms used to identify nearest neighbouring trajectories and how they converge or diverge, the two most popular being from Wolf et al. (1985) and Rosenstein et al. (1993). The latter was chosen for analysis of the data presented in this thesis as it has demonstrated superior performance with smaller data sets and it is robust to higher amounts of signal noise (Rispens et al., 2014). Although this thesis did not address the comparison between algorithms, it did explore the comparison between 6 and 12-dimensional state space reconstruction techniques. The comparison of state space reconstruction techniques has been studied previously, and it was found that state spaces using positions and velocities (i.e. redundant information) should be avoided but display similar qualitative trends (Gates & Dingwell, 2009). In this thesis, the findings show that both methods predict the FE block to be the least locally stable (greater λ_{max}), while there are differences between state-space reconstruction techniques and the estimation of λ_{max} in the rotation and complex task (Figure 6). So, while a trend exists in the FE block, these techniques do not seem to follow a trend in the rotation and complex blocks. While it is difficult to determine which method is more correct in estimating λ_{max}, the exact method of estimation should be carefully considered. λ_{max} in both methods is estimated as the slope of a line of best fit attached to the average logarithmic divergence (from 0-0.5 cycles) of nearest neighbouring trajectories. It is assumed that the divergence curve from 0-0.5 cycles is relatively linear, and therefore a linear line of best fit would be an accurate representation of the slope (Bruijn et al., 2009a). In this case, the 6D technique generally produced a more linear divergence curve and therefore may have been a more accurate and consistent estimation of λ_{max} (Rosenstein et al., 1993; Figure 7). Therefore, it may be that the different trends between 6D and 12D reconstruction techniques may exist because of the instability of λ_{max} estimations in the 12D technique resulting from non-linear divergence curves. This idea is supported by the absence of a significant difference between rotational and complex movements in the 6D method, which agrees with previous research comparing these three movement tasks (Dupeyron et al., 2013). Findings from Gates & Dingwell (2009) also support the 6D method, as constructing a state space using delay embedding produced less error than a state reconstruction using redundant information. Future research should compare these tasks using the 12D reconstruction technique with an increased sample size to improve confidence in findings. Although there may be arguments for either state space reconstruction strategy being superior, these differences in λ_{max} estimations enforce the importance of using consistent state space reconstruction techniques to quantify LDS reliably between studies.

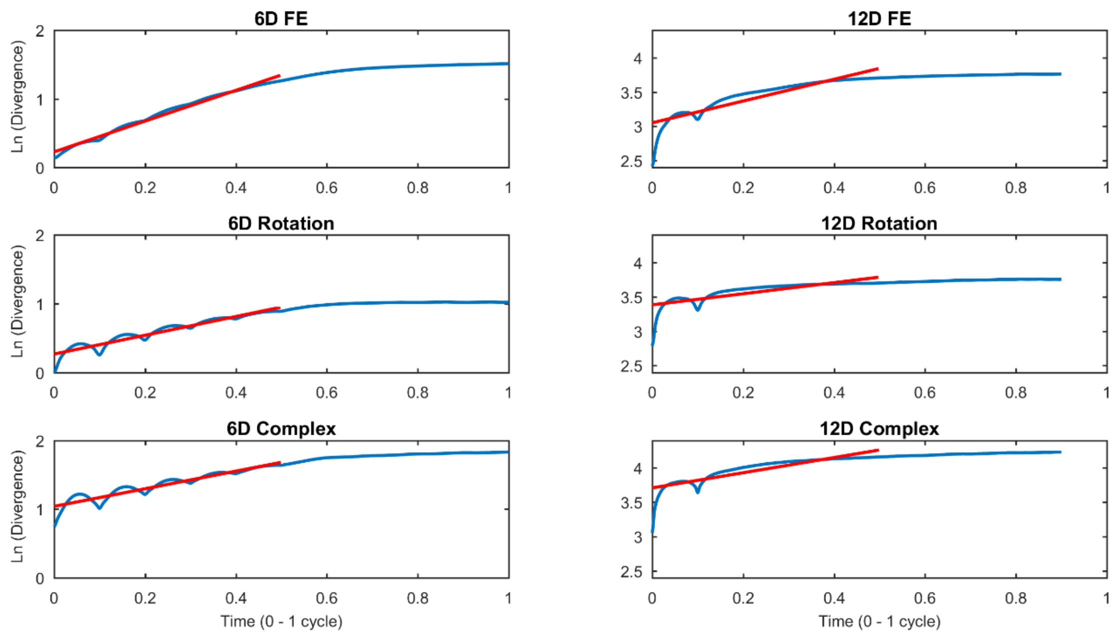

Figure 7. *Average divergence curves using 6D and 12D state space reconstruction techniques for all participants in all conditions. Blue lines represent the rate of divergence, red lines are a linear line of best fit from 0-0.5 cycles. Slope of the red line represents* λ_{max}.

There are several limitations to the current study. First, it is well established that LBP individuals demonstrate different movement patterns compared to healthy controls (van Dieën, Reeves, et al., 2018). As the participants in this study were without LBP, it cannot be assumed that similar trends would be obtained if LBP participants were included in the analysis. Therefore, these techniques may not detect these changes similarly. Additionally, LDS results in one single output that represents the dynamic stability of the spine system (λ_{max}), whereas SI provides many outputs that describe the contribution of different systems. This presents a challenge in how the two methods can be compared. To resolve this issue, a stepwise linear regression was used to construct a regression equation with λ_{max} as the dependent variable and ten outputs of SI as the independent variables to describe the variance in λ_{max}. While this method is an effective technique to find general trends between data sets, it has received criticism due to the limited restriction on the number of independent variables used as predictors. As a result, many independent variables can be included that are unrelated to the dependent variable and the predictive capability of the models could improve strictly by chance. To mitigate this effect, all outputs of SI were considered, and only variables that were believed to be strongly related to movement control were included as independent variables.

CHAPTER 6: CONCLUSION

Overall, the findings of this thesis brought forward two important concepts. Mainly, it demonstrated that using the 6D method of calculating λ_{max} during a repetitive spine rotation task is best described by the motor behaviour outputs of SI in the relax task. Therefore, if LDS were to be used as a clinical screening tool for assessment of motor control, a rotation task should be used if there is potential for further investigation of subsystem contribution using SI. Although the relationship between these variables is strong, with the SI outputs under relax task instructions describing 64.7% of the variance in LDS, there are other comparisons that predict poorly or not at all. This reflects that the changes in neuromuscular control demands from static to dynamic tasks may require large changes in control behaviour that do not allow for findings in either to be interchangeable. This means that although there are some obvious shared aspects of neuromuscular control strategies between these two tasks, careful consideration should be exercised when comparing literature using these techniques. For example, if movement control is being summarized across literature, it would not be accurate to compare λ_{max} values calculated using the 6D method during a complex movement task to outputs of SI. More specifically, as these outputs do not demonstrate linear correlation, it cannot be determined if movement control changes in LDS will be mimicked by motor control changes in SI control outcomes. Future studies should incorporate loaded trunk movements during LDS assessment to increase spine stiffness (Gardner-Morse & Stokes, 2001) and observe if control behaviour is similar to SI in the resist task, where spine stiffness is also increased.

Beyond the differences between LDS and SI outcomes, it is also very important to consider the differences that were observed within LDS between 6D and 12D state space reconstruction methods. While both methods are valid, it is clear that λ_{max} estimations may not follow distinct trends. This means that it may not be accurate to compare literature using different state reconstruction techniques.

Finally, this thesis demonstrates the importance of using consistent techniques to quantify motor control of the spine and to understand the relationship between commonly used quantification techniques. Despite SI and LDS having different advantages, they only capture similar aspects of motor behaviour in very specific conditions. This provides a barrier to summarizing literature using these methods and can contribute to confusion when generalizing movement behaviour within different populations. With a consistent framework for assessing spine motor control, researchers can begin to better understand mechanical risk factors for LBP and how spine control may be related to this common and disabling condition. In addition to laboratory assessment, being able to improve clinical identification of impaired movement behaviour can improve the efficiency of treatment plans. While further investigation is required to determine a reliable system of movement control assessment, this work provides a valuable foundation to guide future research.

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Amsterdam
Movement Sciences

uOttawa

Vrije Universiteit Amsterdam[2]
Amsterdam Movement Sciences

University of Ottawa[1]
School of Human Kinetics

Appendix A

Research Project Title: A comparison of methods to quantify control of the spine

Introduction:
Dear Sir/Madam,

We kindly ask you to participate in an investigation. Participation is voluntary. In order to participate, your written permission is required. Before you decide whether you want to participate (in this research), you will be given an explanation of what the research entails. Read this information carefully and ask the researcher if you have any questions. This research is done by the Faculty of Behavioral and Movement Sciences at the Vrije Universiteit Amsterdam. A total of 30 volunteers will participate in this research. The Standing Committee on Legal Affairs and Ethics of this faculty has assessed this research and concluded that it is within the ethical guidelines of the faculty.

What is the purpose of this research?
Low back pain is an important and common health problem. Some of the patients with low back pain have a problem with controlling body movements and would benefit from improvements in movement control. Recently a new measurement method has been developed to determine whether this problem is present in a patient. In this study, we would like to characterize the movement of the body movement of people without low back pain in different age groups as comparison material for patient measurements. In addition, we want to compare these measurements with the results of an alternative method that is easier to apply in practice. Therefore, we ask you to participate in measurements in which we will characterize the control of your body movements in two ways.

What does participation mean?
If you participate, it takes about 3 hours in total, during this period you perform two tests. In a test (test A) you are on a so-called knee chair and your trunk will be perturbed (this feels as if you are pushed in the back with low force), with the other test (test B) you perform a series of repeated trunk movement tasks. These tasks will be completed in a randomized order.

Measurements
You will visit our laboratory once for a visit of about 3 hours. During the measurements, we ask you to wear your own sportswear (shorts and a t-shirt). Dressing room is available. During the measurements the following will happen:

For test A you are in a half kneeling position on a knee chair. Your pelvis is fixed in a frame so that it cannot move. A pushing rod pushes your back with a small but unpredictable force. This takes about 1 minute and is repeated three times under two different conditions. One condition, you will perform this task while relaxed and allowing the movement of the push rod as much as possible. In the other condition, you will do this while resisting the movement as much as possible. We will measure the movements and the force of the push rod and the activity of your back muscles using electrodes on your back.

For test B you will perform three repetitive movement tasks consisting of 35 cycles to the rhythm of a metronome. The first condition will involve a bending movement of the trunk, touching alternating targets at the height of the shoulder and knee. The second condition will involve repetitive rotational movements of your trunk, and alternating touching targets on your left and right sides at shoulder height. Finally, the third condition will involve repetitive movements that combine a twist and bend between targets at both knees and shoulders. Each condition is practiced briefly in advance and the sequence of the conditions will be determined randomly. We measure the movements of your trunk and pelvis with markers attached to your trunk and pelvis with an elastic band.

What is expected of you?
In order to complete the research, it is important that you comply with the following arrangements:

- The tests are performed according to explanation
- All meetings arranged with the researcher are attended

It is important that you contact the researcher if:

- You are admitted or treated in a hospital
- You experience sudden health issues
- You no longer wish to participate in the research study

Possible risks:
It is highly unlikely that you will experience any adverse effects during these tasks. It is possible that you may feel fatigued during the protocol. If this occurs, please inform the researcher and you will be provided with longer breaks in between trials.

Possible advantages and disadvantages:
Participation in the research will have no direct benefits for you. The results of this research will hopefully prove useful for better diagnostics in people with low back pain. The disadvantage of participation is only the time that it will take to complete the experimental protocol (approximately 3 hours excluding travel time).

If you do not wish to participate
It is solely your decision to participate in this research. Participation is voluntary. If, during the protocol, you wish to cease participation, you may stop immediatley with no explanation necessary. You must, however, inform the researcher that you wish to cease participation. The data collected until that time will be used for analysis. Please notify the researcher of any important

information regarding the procedure, and it will be addresed. You will then be asked if you are willing to continue participation.

Conclusion
The entire research has ended when all participants are finished. After processing all data, the researcher informs you of the most important results of the research. This happens within half a year after your participation. You may be approached for future research in the future. If you do not want this, you can indicate this on the consent statement.

Exclusion criteria:
You will be excluded from participation in this study if you have a history of low back pain. Also if you have previously experienced any: cardiovascular conditions, neurologic disorders (neuropathy, neurodegenerative conditions), low back pain (discogenic, mechanical, myofascial), ankle injury (sprained, fractured) use of medication (anti-inflammatories, analgesics, anticonvulsants, antidepressants), history of low back injury (discogenic mechanical) use of anticoagulent therapy, stroke or TIA, spine trauma, motor vehicle accident, lumbar spine surgery, hypertension, CTD and focal neurological symptoms (sensory/motor) you will be excluded from participation.

Usage and storage of data

For this research, some personal information is required to be collected and used. Each subject will get a code that will appear on the data. Your name will be omitted.

All your information remains confidential. Only the main investigator and the executive researcher know which code you have. The key for the code stays with them. Research reports only use that code. By signing the consent statement, you consent to collecting, storing and understanding your medical and personal information. The researcher stores your data for 15 years.

Compensation for participation
Participation in the research does not cost you anything. You will not be paid for participating in this research. However, you will receive compensation for your (extra) travel expenses. In addition, we would like to thank you with a gift card of € 20 for your participation in the research.

Do you have questions?
For questions, please contact the main investigator.

Thank you for your consideration!

Research Project Title: **A comparison of methods to quantify control of the spine in different age groups**

Consent:
I have read this consent form, and I agree to participate in the procedures of this study.

____
Printed Name of Participant

________________________________ ________________________________
Signature of Participant Date

Investigator Statement (or Person Explaining the Consent):
I have carefully explained to the research participant the nature of the above research study. To the best of my knowledge, the research participant signing this consent form understands the nature, demands, risks and benefits involved in participating in this study. I acknowledge my responsibility for the care and well-being of the above research participant, to respect the rights and wishes of the research participant, and to conduct the study according to applicable Good Clinical Practice guidelines and regulations.

____
Name of Investigator/Delegate (printed)

________________________________ ________________________________
Signature of Investigator/Delegate Date

Informed Consent to have Pictures Taken:
I consent to have side view pictures taken of myself completing the experiment, and understand that no pictures will be taken at any point without me knowing. I also understand that if any of these pictures are used in a subsequent presentation or publication, that my face and any other identifiers will be blurred. You can still participate in the research study without consenting to have pictures taken.

________________________________ ________________________________
Name Date

____
Signature

________________________________ ________________________________
Witness Name Witness Signature

Future Participation:

File Number: H02-17-11 **Date (mm/dd/yyyy):** 03/17/2017

Université d'Ottawa — University of Ottawa

Bureau d'éthique et d'intégrité de la recherche — Office of Research Ethics and Integrity

Ethics Approval Notice

Health Sciences and Science REB

Principal Investigator / Supervisor / Co-investigator(s) / Student(s)

First Name	Last Name	Affiliation	Role
Ryan	Graham	Health Sciences / Others	Supervisor
Eric	Bourdon	Health Sciences / Human Kinetics	Student Researcher
Wantuir Carlos	Ramos Junior	Health Sciences / Human Kinetics	Student Researcher

File Number: H02-17-11

Type of Project: Independent Student Project

Title: The effect of focus of attention on local dynamic stability during repetitive spine flexion

Approval Date (mm/dd/yyyy)	Expiry Date (mm/dd/yyyy)	Approval Type
03/17/2017	03/16/2018	Approval

Special Conditions / Comments:
N/A

File Number: H02-17-11

Date (mm/dd/yyyy): 03/17/2017

Université d'Ottawa
Bureau d'éthique et d'intégrité de la recherche

University of Ottawa
Office of Research Ethics and Integrity

This is to confirm that the University of Ottawa Research Ethics Board identified above, which operates in accordance with the Tri-Council Policy Statement (2010) and other applicable laws and regulations in Ontario, has examined and approved the ethics application for the above named research project. Ethics approval is valid for the period indicated above and subject to the conditions listed in the section entitled "Special Conditions / Comments".

During the course of the project, the protocol may not be modified without prior written approval from the REB except when necessary to remove participants from immediate endangerment or when the modification(s) pertain to only administrative or logistical components of the project (e.g., change of telephone number). Investigators must also promptly alert the REB of any changes which increase the risk to participant(s), any changes which considerably affect the conduct of the project, all unanticipated and harmful events that occur, and new information that may negatively affect the conduct of the project and safety of the participant(s). Modifications to the project, including consent and recruitment documentation, should be submitted to the Ethics Office for approval using the "Modification to research project" form available at: https://research.uottawa.ca/ethics/forms.

Please submit an annual report to the Ethics Office four weeks before the above-referenced expiry date to request a renewal of this ethics approval. To close the file, a final report must be submitted. These documents can be found at: https://research.uottawa.ca/ethics/forms.

If you have any questions, please do not hesitate to contact the Ethics Office at extension 5387 or by e-mail at:

Signature:

The following modifications have now been approved:

1. Study design and location: A sub-study will now be conducted at Vrije Universiteit Amsterdam and surface electromyography (sEMG) of 4 muscles will be collected. In addition to the to the repetitive trunk movement task (for which a condition will be added), a protocol will also be added in which the participant will be ventrally perturbed by a linear actuator while seated in an apparatus.

2. Compensation: Participants in the Amsterdam study will be compensated with a 20 euro gift certificate and travel costs will be reimbursed.

3. Research team: Wantuir Carlos Ramos Junior is no longer part of the team. Dr. Japp van Dieen will be the lead in the Netherlands.

4. Documents: Recruitment and consent materials have been revised accordingly.

The modifications are covered under your current approval, which is still valid until March 16, 2018.

All the best,

Proefpersoneninformatie voor deelname aan onderzoek

Vergelijking van methoden om de sturing van rompbewegingen te karakteriseren en de effecten van leeftijd

Inleiding

Geachte heer/mevrouw,

Wij vragen u vriendelijk om mee te doen aan een onderzoek. Meedoen is vrijwillig. Om mee te doen is wel uw schriftelijke toestemming nodig.

Voordat u beslist of u wilt meedoen (aan dit onderzoek), krijgt u uitleg over wat het onderzoek inhoudt. Lees deze informatie rustig door en vraag de onderzoeker uitleg als u vragen heeft. U kunt er ook over praten met uw partner, vrienden of familie.

Dit onderzoek wordt gedaan door de Faculteit der Gedrags- en Bewegingswetenschappen op de Vrije Universiteit Amsterdam. Er doen in totaal 30 gezonden proefpersonen mee aan dit onderzoek. De Vaste Commissie Wetenshap en Ethiek van deze faculteit heeft dit onderzoek beoordeeld en heeft geconcludeerd dat het binnen de ethische richtlijnen van de faculteit valt.

1. Wat is het doel van het onderzoek?

Lage rugpijn is een belangrijk en veel voorkomend gezondheidsprobleem. Een deel van de patiënten met lage rugpijn heeft een probleem met het sturen van de bewegingen van de romp en zou gebaat zijn bij training van deze vaardigheid. Recent is een nieuwe meetmethode ontwikkeld om vast te kunnen stellen of dit probleem bij een patiënt aanwezig is. In dit onderzoek willen we graag de sturing van de rompbeweging van mensen zonder lage rugpijn in verschillende leeftijdsgroepen karakteriseren als vergelijkingsmateriaal voor metingen bij patiënten. Daarnaast willen we deze metingen vergelijken met de resultaten van een alternatieve methode die makkelijker toepasbaar is in de praktijk. Daarom vragen we u om mee te doen aan metingen waarbij we op twee manieren de sturing van uw rompbewegingen zullen karakteriseren.

1. Wat meedoen inhoudt

Als u meedoet, duurt dat in totaal ongeveer 3 uur, tijdens deze periode voert u twee testen uit. Bij een test (test A) zit u op een zogenaamde kniestoel en wordt uw romphouding door een machine verstoord (dit voelt alsof u in de rug wordt geduwd met lage kracht), bij de andere test (test B) voert u een serie herhaalde rompbewegingen uit. Loting bepaalt in welke volgorde u deze testen zult uitvoeren.

Bezoeken en metingen

U komt 1 keer naar ons laboratorium voor een bezoek van ongeveer 3 uur. Tijdens de metingen vragen wij u uw eigen sportkleding (shorts en een t-shirt) te dragen. Kleedruimte is aanwezig. Tijdens de metingen zal het volgende gebeuren:

Wij vragen u eerst een korte vragenlijst met wat vragen over uw gezondheid in te vullen.

Voor test A zit u in een half geknielde houding op een kniestoel. Uw bekken wordt in een frame vastgezet zodat dit niet kan bewegen. Een duwstang duwt u in de rug met een lichte maar onvoorspelbare kracht. Dit duurt circa 1 minuut en wordt driemaal herhaald. Driemaal voert u deze taak uit terwijl u de beweging die de duwstang veroorzaakt zoveel mogelijk toelaat en driemaal doet u dit terwijl u de beweging zoveel mogelijk tegenhoudt. Wij meten de bewegingen en de kracht van de duwstang en de activiteit van uw rugspieren met behulp van plakkers op uw rug.

Voor test B staat u en voert u in drie series 35 maal achter elkaar dezelfde beweging uit op het ritme van een metronoom. In serie a is dat een buigbeweging van de romp waarbij u met uw vingers afwisselend doelen aantikt op schouder- en kniehoogte. In serie b voert u herhaalde draaibewegingen van uw romp uit waarbij u afwisselend doelen aantikt aan uw linker- en rechterzijde op schouderhoogte. Tenslotte voert uw in serie c herhaalde bewegingen uit waarbij een draaiing en buiging combineert tussen doelen ter hoogte van beide knieën en schouders. Elke serie oefent u kort van tevoren even en de volgorde van de series wort door het lot bepaald. Wij meten hierbij de bewegingen van uw romp en bekken met behulp van kleine lampjes die op uw romp worden bevestigd met een elastische band.

2. Wat wordt er van u verwacht

Om het onderzoek goed te doen, is het belangrijk dat u zich aan de volgende afspraken houdt.

De afspraken zijn dat u:

- de test uitvoert volgens de uitleg.
- afspraken voor bezoeken nakomt.

Het is belangrijk dat u contact opneemt met de onderzoeker:

- als u in een ziekenhuis wordt opgenomen of behandeld.
- als u plotseling gezondheidsklachten krijgt.
- als u niet meer wilt meedoen aan het onderzoek.

3. Mogelijke nadelige effecten

Meedoen aan het onderzoek heeft waarschijnlijk geen nadelige effecten. Wel kan het zijn dat u tijdens de taken wat vermoeid raakt. Indien dit gebeurt, meldt u dit dan alstublieft direct aan de onderzoeker. U krijgt dan langere rustpauzes.

4. Mogelijke voor- en nadelen

Deelname aan het onderzoek heeft voor u geen directe voordelen. De uitkomsten van dit onderzoek zullen hopelijk van nut blijken voor een beter diagnostiek bij mensen met lage rugpijn.

Nadeel van meedoen aan het onderzoek is eigenlijk alleen de tijd die het u kost (3 uur exclusief reistijd).

6. Als u niet wilt meedoen of wilt stoppen met het onderzoek

U beslist zelf of u meedoet aan het onderzoek. Deelname is vrijwillig.

Doet u wel mee aan het onderzoek? Dan kunt u zich altijd bedenken. U mag tijdens het onderzoek stoppen. U hoeft niet te zeggen waarom u stopt. Wel moet u dit direct melden aan de onderzoeker. De gegevens die tot dat moment zijn verzameld, worden gebruikt voor het onderzoek.

Als er nieuwe informatie over het onderzoek is die belangrijk voor u is, laat de onderzoeker dit aan u weten. U wordt dan gevraagd of u blijft meedoen.

7. Einde van het onderzoek

Uw deelname aan het onderzoek stopt als

- het aantal bezoeken zoals beschreven onder punt 2 voorbij is,
- u zelf kiest om te stoppen,

- de onderzoeker het beter voor u vindt om te stoppen,
- de ethische toetsingscommissie of de overheid besluit om het onderzoek te stoppen.

Het hele onderzoek is afgelopen als alle deelnemers klaar zijn. Na het verwerken van alle gegevens informeert de onderzoeker u over de belangrijkste uitkomsten van het onderzoek. Dit gebeurt ongeveer binnen een half jaar na uw deelname. Mogelijk wordt u in de toekomst nog benaderd voor vervolgonderzoek. Mocht u dit niet willen dan kunt u dit aangeven op het toestemmingsverklaringsformulier.

8. Gebruik en bewaren van uw gegevens

Voor dit onderzoek is het nodig dat enkele persoonlijke gegevens worden verzameld en gebruikt. Elke proefpersoon krijgt een code die op de gegevens komt te staan. Uw naam wordt weggelaten.

Al uw gegevens blijven vertrouwelijk. Alleen de hoofdonderzoeker en de uitvoerend onderzoeker weten welke code u heeft. De sleutel voor de code blijft bij hen. Ook in rapporten over het onderzoek wordt alleen die code gebruikt. Als u de toestemmingsverklaring ondertekent, geeft u toestemming voor het verzamelen, bewaren en inzien van uw medische en persoonsgegevens. De onderzoeker bewaart uw gegevens 15 jaar.

9. Verzekering voor proefpersonen

Voor iedereen die meedoet aan dit onderzoek is een verzekering afgesloten. De verzekering dekt schade door het onderzoek. Niet alle schade is gedekt. In de bijlage vindt u meer informatie over de verzekering. Daar staat ook aan wie u schade kunt melden.

10. Vergoeding voor meedoen

Deelname aan het onderzoek kost u niets. U wordt niet betaald voor het meedoen aan dit onderzoek. Wel krijgt u een vergoeding voor uw (extra) reiskosten. Daarnaast willen wij u graag bedanken met een cadeaubon van € 20 voor uw deelname aan het onderzoek.

12. Heeft u vragen?

Bij vragen kunt u contact opnemen met de hoofdonderzoeker.

Dank voor uw aandacht.

Bijlage a: informatie mbt verzekering

-English-

The Scientific and Ethical Review Board (VCWE) of the Faculty of Behavior & Movement Sciences, VU University Amsterdam, has reviewed your research proposal entitled "A comparison of two different methods to assess trunk control and the effects of age on trunk control" on ethical aspects.

Based on the submitted information, the board has no major ethical objections. However, there are a few relatively minor remarks on how to improve and/or clarify your protocol. You find them below. *We trust that you will incorporate these as adequately as possible.* With that, the research proposal complies with the ethical guidelines of the faculty, and does not need to be re-assessed by the committee. This positive advice is valid for 5 years after today's date.

We wish you all the best in your research.

Met vriendelijke groet | With kind regards,

Appendix B

Segment	Marker name	Location
Pelvis	rasis	Right anterior superior iliac spine
	lasis	Left anterior superior iliac spine
	rpsis	Right posterior superior iliac spine
	lpsis	Left posterior superior iliac spine
Thorax	T8	8^{th} thoracic vertebrae
	xiphoid	Xiphoid process
	C7	7^{th} cervical vertebrae
	Sternal_notch	Sternal notch