Dynamic stability of the human body during unstable pushups

by

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Abstract

Instability training provides the nervous system with a greater challenge than traditional training, and thus can be performed with a reduced external load, potentially reducing the likelihood of injury. Unstable exercises have also been found to elicit higher levels of core muscle activity than their stable counterparts and so may increase joint loading. The degree to which instability challenges the stability of the human body likely relies on the level of instability of the movement, but the influence of experimental instability, and its effect on the level of stability of the human body, has yet to be determined. This study aims to examine how altering the available degrees of freedom of the pushup, as a means of quantifying instability, may affect the dynamic stability of the lumbar spine. The results of this study indicate that adding two available degrees of freedom to the conventional pushup significantly increases kinematic variance of the L1 vertebra through 3-dimensional space. No significant difference in kinematic variability existed between the two conditions of additional available degrees of freedom. These results may help health care professionals improve the individualization of their training programs by taking desired level of instability into account, and adjusting available degrees of freedom accordingly.

Keywords:

Instability training, dynamic stability, spine stability, degrees of freedom, unstable pushups

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CHAPTER 1

Introduction

1.0 Introduction

Exercises which challenge stability have previously been suggested to be effective in directly decreasing low back pain (Hides, 2001; O'Sullivan, 1997). Presently, research has attempted to quantify spine stability during unstable exercise, but no method exists to define a progressive level of instability of the exercises being used. Exercises which challenge stability are also capable of producing the same amount of muscle activity as their stable counterparts (Vera-Garcia, 2000; Behm, 2005), but with a lower associated external load. This decrease in load applied to the human body suggests a decreased potential for load-induced injury (Behm, 2006), assuming internal load is not increased.

Using motion capture technology, it is possible to model the local stability of the spine in 3-dimensional space and determine kinematic variance over time (Graham, 2012a). Kinematic variance has been used in the past as a means to quantify the dynamic stability of a system through 3-dimensional space (Tanaka 2009, Graham 2012a). Dynamic stability has yet to be quantified across exercises with graduated degrees of freedom (DOF).

The purpose of the present study was two-fold: to determine the feasibility of using available DOF as a measure of instability of an exercise, and to examine how the level of stability of the human body changes with manipulation of the number of available DOF. The hypothesized outcome is that Lyapunov analysis of the pushup will indicate less stability as more DOF are made available.

CHAPTER 2

Review of Literature

2.0 Introduction

Responding to an unstable exercise requires the body to adjust its position constantly (Cosio-Lima, 2003) and to coordinate both synergistic and antagonistic muscles to a greater degree than traditional methods of training (Behm, 2006). Unstable exercises have been shown to increase the activation of not only limb musculature but of the core musculature as well, when compared to stable variations of the same movement (Vera-Garcia, 2000; Behm, 2005). Unstable exercise has even been shown to directly improve pain scores in low back pain populations (O'Sullivan, 1997; Hides, 2001).

Stability can be defined as a system's ability to survive or resist an applied perturbation (Grenier, 2007), and dynamic stability describes a system's ability to stabilize over a given period of time. In recent years, several studies have proposed and tested a variety of methods aimed at quantifying dynamic stability (England, 2007; Tanaka, 2009; Graham, 2012a). Quantifying the kinematic variance of a vertebrae of interest through motion capture analysis has been shown to be a valid and reliable method of estimating dynamic stability of the human body (Graham 2012a, 2012b).

While instability challenges the neuromuscular system, it is unclear as of yet if a threshold of excessive instability exists. It is conceivable that the level of instability of an exercise may influence the degree to which dynamic stability of the body is challenged or compromised. As an exercise gets more unstable, there may be a tipping point at which the participant may be unable to regain stability, thereby leading to injury. Importantly, the effect that manipulating the available DOF of an exercise has on dynamic stability of the participant has not yet been determined.

2.1 Overview Of Spinal Anatomy, Stability And Pathology

2.1.1 Vertebral Anatomy and Function

The spine is a complex network of bones, muscles and connective tissue which interact with each other to provide the skeleton with structure and movement. At birth the vertebral column is made up of 33 vertebrae which can be divided into five sections: cervical, thoracic, lumbar, sacral and coccygeal. As an infant develops, the four bones of the coccygeal region fuse together and the five bones of the sacral region also undergo fusion. This results in a reduction to twenty-six vertebrae upon complete development. The vertebrae of the cervical, thoracic and lumbar regions all share similar properties in terms of general structure. One will notice, however, an increase in vertebral size as the spine is examined from cranial to caudal, corresponding to the increase in weight that the vertebrae need to support (Kreighbaum, 1996). The spine serves to transmit loads from the trunk to the lower limbs, to protect the spinal cord, and serves as a point of attachment for the axial muscles and the ribs (Marieb, 2010). A normal spine has four regions of curvature, which correspond to the different vertebral regions. The cervical and lumbar regions are concave in the posterior direction while the thoracic and sacral regions are concave anteriorly. These curvatures play an important role in giving the spine its spring-like property and flexibility. The spine also utilizes a unique combination of active and passive dynamics to maintain stability and structure (Panjabi 1992).

2.1.2 The Passive System

The passive system of the spine consists of intervertebral discs and ligaments. Between the bones of the spine are the intervertebral discs, which allow a small amount of movement between vertebrae, as well as acting as shock absorbers to disperse impact forces (Kreighbaum, 1996). These intervertebral discs are composed of two sections – a gelatinous interior area which allows the disc to move, surrounded by a network of collagen and fibrocartilage to prevent excessive movement of the soft interior. The fibrous area is directly attached to the two vertebrae above and below it, thus preventing the vertebrae from excessive rotation or shear forces (Urban, 2003). Within the discs are proprioceptors, which provide feedback regarding the current position and degree of movement of their respective vertebrae (Holm, 2002). The intervertebral discs are subject to compression anytime we are in an upright position, and are subject to high compressive forces when we take an external load (Haff, 2015). Only problematic when surpassing the tolerance of the spine, these compressive forces are actually required to maintain spinal stability (Willardson, 2007). If no compressive forces existed upon the vertebrae and intervertebral discs, the stability of the system would be greatly compromised, similar to a tall stack of dominos. Along with these intervertebral discs, intervertebral ligaments exist which are capable of preventing some displacement in any given direction. Underactive musculature, however, may result in excessive reliance on the ligaments of the vertebral column. Under high loads, this creates the potential for damage to the vertebrae, the intervertebral discs, the muscles and the ligaments themselves.

2.1.3 The Active System

The lumbar vertebrae, due to their caudal position and proximity to the pelvic girdle, receive the most stress of all vertebral regions (Brinckmann 1989). When unsupported by the active system, the osteoligamentous lumbar spine has been found to buckle at a load of just 88 N (Crisco, 1992). For context, this is equivalent to a load of just under 20 pounds. To help maintain stability, several core muscles cross the area in order to assist the intervertebral discs and ligaments in maintaining stability. These muscles which contribute to spinal stability can be referred to as the active system. To prevent buckling of the spine every time a load is taken, these muscles are activated to stiffen the spine and subsequently increase stability (Carpenter, 2001; Kohler, 2010). Similar to the guy wires holding up a telephone pole, activation of the core musculature contributes to the maintenance of spinal stability by increasing functional compression of the spinal column (Grenier, 2007). The posterior muscles of note that cross the lumbar region of the spine are the multifidus, quadratus lumborum and iliocostalis lumborum. The anterior muscles of note in this area are the rectus abdominis, external oblique, internal oblique and transversus abdominis. These muscles act as prime movers during movement and exercise, and can also provide stability across several vertebral segments simultaneously (Behm, 2010).

The transversus abdominis specifically has been the focus of many recent studies. In the pursuit of de-constructing the effect of core musculature activation on spinal stability, one study suggested the transversus abdominis may reduce the loading of the lumbar spine through an increase in intra-abdominal pressure (Cresswell, 1992). The

transversus abdominis has also been found to be the first muscle to activate when a perturbation was applied to the trunk unexpectedly (Cresswell, 1994). While the transversus abdominis has been suggested to play a role in spinal stability, it was more recently discovered that activating the transversus abdominis in an isolated manner is virtually impossible (Vezina, 2000). In a comparison of muscle activity across all major core muscles, the activation of all different abdominal muscles increased in relation to exercise intensity, such that at high levels of intensity every examined muscle was activated to a high degree (Davidson, 2005).

These studies suggest that in any situation where the equilibrium of the spinal system is compromised, a precise coordination of the passive and active systems is required. The feedback from the spinal proprioceptors and the control of the appropriate muscles must be synchronized to adapt to the stimulus and regain adequate stability. The efficiency of this system can be improved with appropriate training, which will be discussed in further sections.

2.1.4 Low Back Pain

Weakness of the core and lower back musculature, along with poor flexibility and lumbar spine instability, are some of the physical factors related to development of chronic low back pain (Katch, 2011). As one ages, the natural tendency for weight gain in the abdomen and the tendency to neglect development of the core musculature combine to make anterior pelvic tilt a common problem (Kreighbaum, 1996). This anterior pelvic tilt is also commonly referred to as lordotic posture. In lordotic posture the abdominals, hip extensors, gluteal muscles and hamstrings become lax, while the back extensors and

hip flexors are shortened and tight (Kreighbaum, 1996). These biomechanical adaptations result in compression of the posterior side of the intervertebral discs and stretching of the anterior vertebral ligaments, and if left unchecked may result in accentuated low back pain (Kreighbaum, 1996). An increase in rectus femoris activation as a hip flexor to compromise for weak abdominals, for example, has been suggested to increase low back pain risk (Youdas, 2008) by contributing to lordotic posture (Jakobsen, 2012). While the results of some studies infer that the transversus abdominis may be less critical than previously believed (Vezina 2000, Davidson 2005), a study of low back pain subjects found a delay in transversus abdominis activation in some (Hodges, 1999). This late activation of the transversus abdominis may be an indicating factor for a spine which is in a chronic state of instability (Grenier, 2007).

A deficiency in the endurance of the low back musculature has also been found to have a strong association with low back pain (Nourbakhsh, 2002; McGill, 2003). It has been previously demonstrated that continuous contraction of only moderate magnitude of the core muscles is sufficient in maintaining stability in a broad sample population ranging from clinical patients to athletes (Cholewicki, 1996). Based on this low strength requirement, the combination of neuromuscular control and muscular endurance are suggested to be the major factors contributing to lumbar instability and resultant low back pain (McGill, 2003). The passive system may also contribute to the development of low back pain. While joint mobility is often beneficial to range of motion and musculoskeletal health, joint laxity due to excessively flexible ligaments may lead to unstable motion when a load is applied (McGill, 2003) which could lead to various pathologies and pain in the lower back.

From a neuromuscular control standpoint, it has been found that many sufferers of low back pain have deficiencies in postural control and lumbar positioning (Parkhurst, 1994, Radebold 2001). What has not been confirmed is if these deficiencies are causes of low back pain, or if tissue damage which caused the low back pain also caused these issues (McGill, 2003). Upon a sudden perturbation of spinal stability, the majority of people should have enough strength to maintain stability (Cholewicki 1996). Thus, it is suggested that low back instability and pain comes from low muscular endurance or a deficiency in neuromuscular control, where a mis-timed muscle activation may be the culprit (Radebold, 2000). Another cause of instability may be an impaired proprioceptive or sensory system from previous injuries. Considering feedback from local sensory organs in the spine is important to postural control and lumbar positioning, it is conceivable that a previous injury in this area may hinder the ability of the neuromuscular system to respond effectively. The source of the control mechanism must also be considered when analyzing the stability of the spinal system. Performing a voluntary movement uses a different control system than responding to an unexpected stimulus. Voluntary movements may arise from an endogenous origin, while reactionary movements may arise in response to an exogenous source (Bouisset, 2008), following different pathways. While muscle activity patterns as a result of the different control mechanisms may be similar, it would be ill-advised to consider action and reaction stabilization as the same concept.

Low back pain is not just present in sedentary populations, but in athletic populations as well. During exercise, the core muscles work to stiffen the intervertebral joints as previously discussed. The combination of this effort to stabilize with the added

external load can create environments of very high compressive forces in the lumbar region; someone who weighs 90 kg and squats 315 lbs can experience peak compression forces of over 13,000 N (Katch, 2011). In national level power lifters, compressive forces of 17,000 N on the L4-L5 joint have been measured (Katch, 2011). In reference, forces of 3400 N are at the recommended limit for compression of the spine as set by NIOSH guidelines (Elfeituri, 2002). As these compressive forces increase, an increase in low back muscle stiffness also occurs, which lends itself to the stability of the individual lumbar intervertebral joints (Brown, 2010). As the load and muscle force both increase, the stiffening effect has been found to plateau, which may lead to impaired stability control of the spine and in turn lower back injuries (Brown, 2008).

2.1.5 Quantifying Spinal Stability

The stability of the spinal column is of utmost importance in tasks involving load transfer and to avoid pain. Spinal stability impairment may result in uncontrolled movement, hindered transfer of forces and damage to the bones and surrounding tissues (Cholewicki, 1996; Graham, 2012b). Definitions of spinal stability has been changing and evolving with the literature published about the topic. A simple yet comprehensive definition described static spinal stability as “the ability of the spinal column to survive an applied perturbation” where, if the “[perturbation] is greater than the potential energy of the column, equilibrium will not be regained” (Cholewicki, 1996). A topic of importance discussed in this study is the difference between excessive movement and instability. “Excessive motion does not imply instability, only the potential for instability” (Grenier, 2007).

A definitive, limitation-free method of quantifying spinal stability has yet to be established (Graham, 2012b). The potential energy of the lumbar system has been measured in recent years in static situations (Bergmark, 1995; Cholewicki, 1996; Grenier, 2007), which comes from contributions of both the active and passive systems of the spine. Considering the passive system to be five rigid segments separated by torsional springs, and the active system to consist of linear springs representing the muscles, the work done by the load and the work done by the system can be used to calculate a spinal stability index at any given point using collected kinetic and EMG data (Grenier, 2007). These static studies examining instantaneous spinal stability opened the door to incorporating the effect of time into the analysis.

In dynamic spinal stability analysis, the stability of the spine through time is considered, which has been used to analyze spinal stability while seated on an unstable chair (Tanaka 2009), throughout a shift, or during a repetitive workload (Graham 2012a). Collected EMG and kinetic data may be combined with passive contributions to generate a rotational stiffness value for each sample, then averaged over time before being input into a further Lyapunov analysis (Graham, 2012b). McGill et al. (2013) analyzed spinal loads during unstable pushing exercises, also using EMG and kinetic data to contribute to an estimate of spinal loading. Kinetic data produced a 3D model of the body performing the movements, which was input into a model to generate values for the restorative properties of the ligaments and intervertebral discs. Normalized EMG and length-velocity properties of the muscles through the movement were combined with these passive contributions to determine L4-L5 compression and shear forces (McGill, 2013). Another study of time-dependent spinal stability by Graham found that increases in lifting load

and rate increased stiffness of the spinal column (Graham, 2012b). It should be noted, however, that stiffness is only one of the components of spine stability, alongside relative muscle force and displacement (Potvin, 2005). In this study, the authors indicate that stability could not be calculated until static equilibrium was achieved (Potvin, 2005), thus dynamic stability analyses often must include additional information such as magnitude of displacement and a temporal measure.

2.2 Overview Of Training Physiology

2.2.1 Physiologic Effects of Resistance Training

Resistance training is an umbrella term that describes any form of fitness improvement techniques involving the control of a load, be it external or one's own body. Resistance training results in an increase in force production capabilities through muscular and neural adaptations. The most noticeable muscular adaptation is muscular hypertrophy, where an increase in force development stimulates skeletal muscle growth (Katch, 2011). Hypertrophy has been identified in some studies after as little as three weeks of resistance training (Seynnes, 2007), however this was most likely due to the untrained nature of the individuals. A number of factors influence muscle hypertrophy, including resistance load (Schoenfeld, 2013; Schoenfeld, 2015), nutrition and supplementation (Naclerio, 2016, Hyde 2016) and frequency (Dankel, 2016).

Physiologically, hypertrophy involves an increase in the diameter of myofibrils within the muscle fibers, which compounds to result in an increase in muscle cross sectional area (Haff, 2015). During resistance training, both type I and type II muscle fibers may be recruited, although type II fibers implement hypertrophic processes more readily than type I fibers (Haff, 2015). The inclusion of anaerobic exercises to target these muscle fibers should result in a greater increase in muscle size than slow aerobic movements. Unfortunately, hypertrophic training utilized by bodybuilders typically targets the large prime movers of the trained movement, commonly neglecting the smaller stabilizers of the joint (Borreani, 2015) by reducing the number of available DOF. This is commonly done with machines which limit movement of the target joint to

a single plane. Genetics also play a role; persons with a higher genetic proportion of fast twitch muscles may have a greater ability to increase muscle cross sectional area, so long as the program targets these type II fibers appropriately (Haff, 2015). Besides muscular hypertrophy, some muscular adaptations to anaerobic exercise include an increase in sarcoplasmic reticulum and T-tubule density (Always, 1989) and enhanced calcium release (Ørtenblad, 2000).

While the hypertrophic adaptation following resistance training is of importance, an even more powerful adaptation is seen by the nervous system. Neuromuscular adaptations to training have been found to occur before any measurable muscular adaptations occur (Sale, 2008; Behm, 2006). When a new exercise is being learned, or the force requirement of an existing exercise increases, activity in the primary motor cortex of the brain increases concurrently, resulting in increased neurological signal sent to the required muscles (Haff, 2015). In studies of the neurological adaptations of training, the maximal recruitment of fast-twitch muscle fibers specifically was found to be less limited in a trained sample of participants compared to an untrained sample (Aagaard, 2002; Pensini 2002). One study even found that only 71% of muscle tissue was activated in untrained populations performing a maximal effort movement (Adams, 1993).

Along with the central nervous system adaptations, the peripheral motor units also experience changes in response to resistance training. In trained individuals, motor units experience an increased firing rate, with subsequent action potentials being sent before the muscle fiber(s) have time to relax. The overlapping action potentials result increased force production, which when utilized appropriately can lead to an increase in movement strength (Aagaard, 2003). The combination of the previously discussed increase in

recruitment with this increase in firing rate results in a great potential for strength gains (Haff, 2015). It has been suggested that neural adaptations are more important in the early phases of resistance training than muscular adaptations (Hakkinen, 1996). Over the first two weeks of resistance training, neural and muscular adaptations contribute to 90% and 10% of strength gains respectively, with a shift occurring over time that sees muscular adaptations contribute to the majority of gains toward the sixth and eighth weeks (Katch, 2011).

At the smallest level, the neuromuscular system adapts by changing how the motor neurons interact with the muscle fibers. While the study of the neuromuscular junction is difficult to conduct with human subjects, some animal studies have been performed and have found an increase in total terminal branch lengths (Deschenes, 1993), greater end plate area, and number of ACh receptors in the sarcolemma (Deschenes, 2000). These adaptations infer enhanced release of ACh by the motor neuron with an improved ability to receive it in the respective motor unit, which could result in a more efficient motor unit activation.

As the human body ages, several physiologic changes contribute to hinder strength, balance and mobility. Without proper resistance training, aging populations may see a decrease in “general strength” up to 30%, a reduction in joint flexibility, a decrease in bone mineral density and a decrease in neuromuscular performance (Katch, 2011, Haff, 2015). While these changes are difficult to combat, regular training may help slow down or even reverse the natural degradation of the body systems. Several studies have shown the ability for older populations to develop strength (Fiatarone, 1990; Christmas, 2000; Lopopolo, 2006), with many significant responses coming after only eight to

twelve weeks. Yoga (Farinatti, 2014) and swiss ball training programs (Seo, 2012) have also demonstrated improved flexibility measures in elderly populations. Resistance training has the capability to improve not only bone health and flexibility, but can also improve balance and overall strength (Layne, 1999), making it very beneficial to older populations.

Whether considering the hypertrophic adaptation, the neuromuscular responses, or its ability to combat the natural deterioration that comes with aging, resistance training should be considered an important piece of any training program.

2.3 Instability Training And Spine Health

2.3.1 Defining Instability Training

Instability training has been gaining ground in the industry of athlete development due to its ability to create environments which more closely mimic the demands of the athlete's respective sport (Behm, 2002; Behm, 2010; Kohler, 2010). Instability training includes any facet of training which aims to improve balance and coordination through creating an environment to challenge equilibrium. This environment of instability can be generated through altering the base of support or using different tools and devices (Anderson, 2004). A common form of instability training that has been around for decades in gyms worldwide is the use of free-weights as opposed to machines for strengthening purposes. While free-weights such as dumbbells and kettlebells are able to create an environment of moderate instability, new developments and inventions have strived to push instability to new levels.

2.3.2 Effects of Instability Training

Along with creating a sport-specific environment, a strong rationale for instability training is the enhanced neuromuscular control required to move efficiently and effectively (Behm, 2006). The coordination requirement imposed by instability training will stress the neuromuscular system to a greater degree than traditional methods as it must control synergist and antagonist muscles to increase joint stability, as well as the agonist muscles required for movement (Behm, 2006). When performing unstable exercises, the continuous challenge to balance and posture results in constant adjustment

of the body to prevent toppling over (Cosio-Lima, 2003). These constant adjustments require the feedback of sensory receptors to relay the current position of limbs and joints, the central nervous system to determine the correct course of action and the response of the small stabilizer muscles to correct for the perturbation. Training this feedback loop should make the entire system more efficient, and may prepare the individual for an unstable environment in everyday life. Besides training the neuromuscular system, a secondary application of instability training is to enhance trunk and core muscle fitness. Several studies highlight the effectiveness of instability devices in a laboratory setting. Unilateral upper body exercises (Behm, 2005) and curl-ups (Vera-Garcia, 2000) have been found to induce greater trunk stabilizer muscle activity than stable versions of the same movements. In a study comparing pushup variations, it was reported that performing pushups on an unstable surface increased both limb and core activation (Anderson, 2003) in an effort to stabilize the system. In short, training with instability is effective to train the target muscles while incorporating core stabilizer muscles, however the degree to which this occurs should be considered before engaging in or prescribing the activity.

The effects of instability training are not just limited to the core. As mentioned, performing an unstable movement requires precise coordination of antagonist and synergist muscles as well as the agonist, a phenomenon called co-contraction. This simultaneous control of muscles on either side of the joint creates a training effect of enhanced dynamic stability throughout the motion in question (Haff, 2015). The gluteus maximus has been shown to be an active stabilizer of the walking lunge (Alkjaer, 2012), and the serratus anterior was shown to increase in activity in the unstable pushup

compared to the stable variant (Borreani, 2015). Training these muscles to stabilize their respective joint could prove important in preventing instability during the various movements of daily life.

2.3.3 Considerations for Instability Training

While instability training is beneficial to increase core activation, stabilizer activation and create a more sport specific environment, it may not be suitable for all populations or applications. The primary argument against instability training for some is the decrease in external force production as a by-product of the increased co-activation (Behm, 2006; Behm, 2010; Kohler, 2010). In fact, a decreased force production of almost 60% was found in the unstable chest press in comparison to the stable variant (Anderson, 2005). This may be due to the contribution of the antagonist muscles; the stability of the joint increases, but activation of the antagonist as a stabilizer may result in increased resistance against the agonist muscle (Behm, 2010). There is also a decrease in the rate of force production and movement velocity. Under unstable conditions, the rate of force development drops by as much as 70% compared to the same movement performed in a stable environment (Drinkwater, 2007).

Many studies regarding the difference between unstable and stable exercises typically involve individuals with little to no experience with instability training (Kohler, 2010). When resistance-trained participants were studied, moderately unstable bases provided no increase in muscle activity compared to their stable counterparts (Wahl, 2008). Devices creating instability could elicit a response of increased core and stabilizer muscle activity in a trained population, although the degree to which this occurs may be

dependent on the level of exercise instability (Wahl, 2008). This was the first study to highlight that low to moderate levels of instability may not be enough to sufficiently challenge the neuromuscular pathways of highly experienced resistance-trained persons.

While not yet studied, it is conceivable that a relationship may exist between the progressive potential for instability of an exercise and the degree to which it challenges the dynamic stability of the human body performing the exercise. In a vulnerable population, such as individuals with compromised core strength or an injury, the potential for unstable movement to challenge the core musculature and spinal stability system with a number of available DOF that is above the individual's tolerance does exist and should be considered carefully. This excessive challenge could prove damaging to the musculoskeletal system of the individual if not implemented in slow progressive steps.

2.3.4 Effects of Instability Training on Spine Health

Instability exercises have been shown to be effective in directly decreasing low back pain (Hides, 2001; O'Sullivan, 1997), although the source of this effect has yet to be discovered (Grenier, 2007). As discussed, one function of the abdominal core muscles is to increase pressure in the abdomen and functional spinal compression to stabilize the lumbar spine. Considering how unstable exercises can increase activation of the core muscles as compared to more stable exercises, such as dumbbell exercises versus barbell exercises respectively (Behm, 2010), it is conceivable that this increased core recruitment may result in core strength gains that are directly applicable to maintaining spinal stability.

Another aspect of spinal stability is the timing of core muscle recruitment. In a study examining trunk muscle activity following an unexpected perturbation, participants with chronic low back pain exhibited a delayed response of the core muscles to said perturbation (Radebold, 2000). These researchers expanded on this finding and discovered this delayed response to be correlated with poor postural control (Radebold, 2001), which may place the lumbar spine at risk (Cholewicki, 2005).

An increase in magnitude of core recruitment as a stabilizer has been found in unstable varieties of chest presses (Behm, 2005), pushups (Holtzmann, 2004) and curl-ups (Vera-Garcia, 2000). When applied as a rehabilitation tool following careful consideration of individual needs and risks, the use of unstable devices to reduce low back pain has been proven effective (Behm, 2010).

Instability training may also play a role in improving the timing of spinal reflexes in an attempt to mitigate risk of low back injury. Performing activities that require neuromuscular control via spinal reflexes may improve this reflexive control of joint positioning following a sudden perturbation (Griffin, 2003). At present, a gap in the literature exists regarding spinal reflexes following an instability training program. A systematic review of the literature pertaining to spinal reflexes and balance training suggests that performing balance training under various conditions (eyes open, eyes closed, one foot) should result in improvements of postural and neuromuscular control in the lower limb (Zech, 2010). While these studies examined standing balance, it is conceivable that performing unstable exercises targeting the muscles of the upper body and core may have a similar effect in improving the latency of core muscle onset following unexpected perturbations.

The core muscles play an important role in stabilizing the spine during exercise, and deficiencies in the strength and/or timing of these muscles may place the participant at risk of injury. These properties of the core muscles may be improved through training on unstable devices, however there is a gap in the literature in terms of quantifying the level of instability during said exercises. What if the exercise is simply too unstable for the participant to handle?

CHAPTER 3

Methods

3.1 Participants

10 resistance trained volunteers (n=6 males, 4 females; age = mean 22.7, SD 1.42; height = mean 175.9 cm, SD 11.99; mass = mean 76.2 kg, SD 11.91) participated in this study. Participants were recruited through poster advertisements placed in the University Athletics Center and through word-of-mouth. The participants self-reported to be free of low back pain, free of current acute musculoskeletal conditions of the hip, knee, shoulder and spine, free of hernia (both sports-induced and otherwise), free of chronic hypertension, to not have been pregnant within 6 months of the study, and to not have had surgery of any type within 6 months of the study. Participants were advised upon initial contact to refrain from resistance training within 48 hours of participating in this study. This study was approved by the University Research and Ethics board, and each participant provided informed consent prior to their participation.

3.2 Procedures

The protocol was completed in single sessions of approximately 90 minutes duration. Fixation of electromyographic equipment and LED (light-emitting diode) triads began immediately following the completion of the consent form. The intercristal line was marked with a ball point pen upon palpation of the iliac crests (Chakraverty, 2007). Following this line medially to the spinal column allowed for a mark to be placed on the L3 spinous process in females and the L4 spinous process in males (Chakraverty, 2007) to accommodate for the anatomical differences in pelvic structure and orientation. Upon palpation of the spinous process of the respective vertebra, the processes of the L1 and L5

vertebrae were palpated through counting vertebrae up and down, respectively. An LED triad was placed on each of these vertebra of the participant, with the middle LED of the triad directly over the spinous processes of the respective vertebrae, as shown in Figure 1.

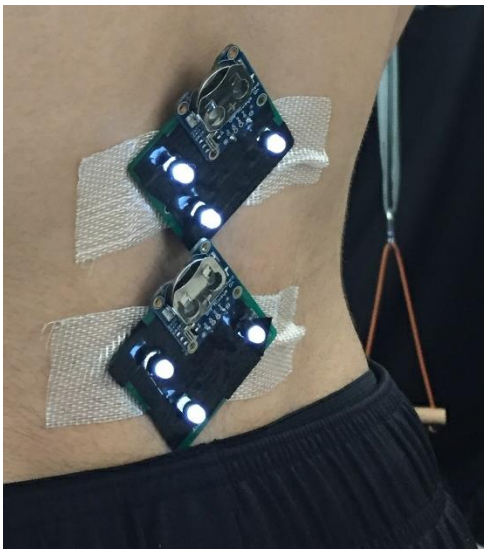

Figure 1: LED Triad and Placement

EMG electrodes were placed bilaterally on the External Oblique (EO), Rectus Abdominis (RA), Upper Erector Spinae (ES) and Longissimus in the area of the multifidus (LG) muscles following the orientation of the muscle fibers. Reference electrodes were placed on the anterior superior iliac spine (Galen, 2015). Specific placement of the electrode pairs is as follows: EO – 3 cm lateral to the semilunar line of the abdomen; RA – 3 cm lateral to the navel; ES – 3 cm lateral to the spinous process of the T9 vertebra (McGill, 2014); LG – 3 cm lateral to the spinous process of the L3 vertebra (Stokes, 2003). Upon fixation of the electrodes, a maximum voluntary isometric contraction (MVIC) of the core musculature was performed and EMG data was collected. The MVIC protocol for the anterior core musculature consisted of a curl-up against manual resistance, while the MVIC protocol for the posterior musculature consisted of a back extension against manual resistance (Konrad, 2005).

Following fixation of the LED triads and electrodes, the parallettes to be used in the baseline condition were placed in the experimental area, and the participant was asked to perform one pushup in the field of view of the cameras. The positioning of the cameras

was adjusted to ensure both LED triads were in the field of view of all three cameras throughout the entire pushup range of motion.

One LED on the triad over the L1 vertebra was used to estimate dynamic stability of the entire human body system. The initial procedural design of this study involved analyzing the movement of two LED triads relative to one another, one on the L1 vertebral body and one on the L5 vertebral body. Preliminary analysis of the movement of these markers relative to one another demonstrated extremely small movements between the L1 and L5 vertebrae; these deviations were no larger than the margin of error in the digitization process, thus we were unable to detect these movements. The back-up plan of estimating dynamic stability of the torso during the pushup was employed. The present chapter describes a method of tracking gross movement of the human body through 3D space during pushups of various levels of instability, tracking the vertebral body of the L1 vertebra as a central point.

3.3 Instrumentation

Muscle activity was collected using an 8-channel wireless EMG system (OpenBCI, Brooklyn, NY) at a sampling rate of 1000 Hz. Electrodes (3M Red Dot, 3M, St. Paul, MN) were affixed bilaterally to the skin following orientation of the muscle fibers of interest, in accordance with the literature (McGill, 2014). The muscles of interest were the external obliques (EO), rectus abdominis (RA), upper erector spinae (ES), and longissimus (LG) in the area of the multifidus (Stokes, 2003). The L1 and L5 vertebrae were marked with LED triads constructed by a member of the research team

from individual parts. Stable pushups were performed on 14” steel parallettes (Rogue Fitness, Columbus, OH). Experimental trials were performed on the Yoak trainer (YOAK Inc., Sudbury, ON) which was used to manipulate the number of available DOF. Three HD action cameras (Yi Technology, Bellevue, WA) were used to collect video recordings of all trials at a frame rate of 60 frames per second (fps) and a resolution of 2.7K. Open source processing software (Argus, UNC-Chapel Hill, Chapel Hill, NC) was used for all calibration and processing of collected footage.

3.4 Camera Synchronization

To synchronize the three cameras, any and all video captures contained three claps at the beginning of the video. Open-source software (Argus) was used to synchronize the audio spikes arising from these three claps; the software analyzed the audio signals and determined the offset of the three spikes of audio signal from each video recording. The three video recordings were then shifted along the time spectrum by this number of frames to synchronize all three video recordings in regards to time.

3.5 Camera Calibration

Upon positioning of the cameras in relation to the area of interest, and with the participant now off to one side, calibration videos were recorded to be used in future processing. A calibration wand of 20 cm

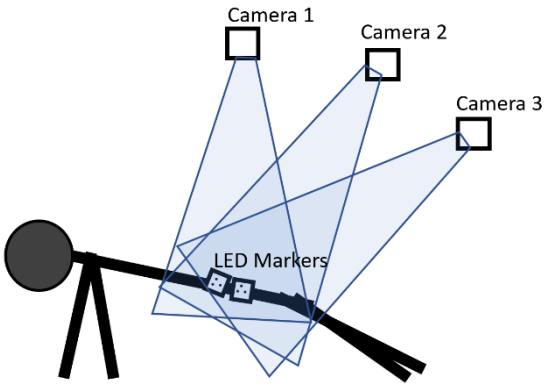

Figure 2: Schematic of camera position in relation to subject

was moved fluidly through a space just larger than the area of interest to ensure the movement of the LED markers would always be within the calibrated area. Upon digitization of the calibration wand, an output file was created by Argus consisting of the resultant 3D coordinates of each end of the wand. The distance between the two points was calculated to determine the error of the digitization/calibration process. All calibrations resulted in an average calculated length of 20.0 ± 0.2 cm. Further explanations of the 3D transformation procedures implemented in the Argus software can be found in Theriault, 2014.

3.6 Experimental Conditions

Four experimental conditions of different DOF were designed for this study, with each participant performing three of them. Each condition involved performing 5 pushups. The number of available DOF was manipulated between conditions. Prior to performing the experimental procedure, participants undertook a warm-up consisting of muscle specific activation drills and pushups regressed to a low intensity. The participants also received

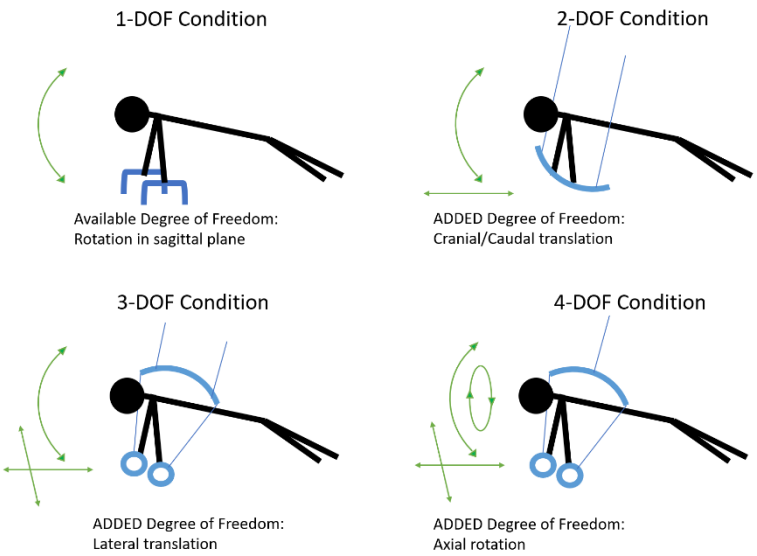

Figure 3: Available degrees of freedom for each experimental condition

guidance pertaining to proper pushup technique. All warm-ups, instruction, and experimental procedures were performed under supervision.

Each participant performed pushups under 3 of the 4 experimental conditions. The different conditions were 1-DOF, 2-DOF, 3-DOF, and 4-DOF. All participants performed the 1-DOF condition, then attempted the 3-DOF condition before advancing to the 4-DOF condition or regressing to the 2-DOF condition. Once three sets of pushups in different conditions were completed, they were labelled Stable, Level 1, and Level 2, regardless of which DOF conditions were employed (figure 4).

The 1-DOF condition (conventional pushup) was performed by all participants. This condition consisted of pushups performed on fixed paralettes. The available degree of freedom was rotation of the body in the sagittal plane about the fulcrum of the feet.

Participants then performed the 3-DOF condition. This condition consisted of pushups performed on two gymnastics rings suspended from the Yoak, in turn suspended from the ceiling by two anchor points. The rings were suspended at the level of the paralettes to ensure consistency in hand height, and thus load, between trials. The available DOF were rotation about the feet in the sagittal plane, translation of the hands in the cranial/caudal direction, and translation of the hands in the lateral direction.

Participants who were able to execute the 3-DOF condition with adequate ability based on subjective analysis by the supervising strength and conditioning specialist were graduated to the 4-DOF condition. Participants who struggled to complete the 3-DOF condition were regressed to the 2-DOF condition. This judgement was a subjective decision made by the supervising Certified Strength and Conditioning Specialist (CSCS)

member of the research team. While the particular conditions would thus be different across participants, the collected data still consisted of a 1-DOF pushup and two sets of pushups at a higher number of available DOF.

The 2-DOF condition consisted of pushups performed on the Yoak suspended from the ceiling by two anchor points. The Yoak was suspended at the level of the paralettes to ensure consistency in hand height. The available DOF were rotation about the feet in the sagittal plane and translation of the hands in the cranial/caudal direction.

The 4-DOF condition consisted of pushups performed on rings suspended from the Yoak, which was in turn suspended from the ceiling by one central anchor point. The rings were suspended at the level of the paralettes to ensure consistency in hand height. The available DOF were rotation about the feet in the sagittal plane, translation of the hands in the cranial/caudal direction, translation of the hands in the lateral direction, and axial rotation of the body. For discussion purposes, the different conditions performed by each participant were considered, based on a relative level of difficulty, Level 1, Level 2, and Level 3. Five participants (1 M, 4 F) were regressed to conditions 1, 2 and 3, and five participants (5 M) performed conditions 1, 3 and 4.

3.7 Protocol

Between the warm-up and the first experimental trial, and again between all subsequent trials, a minimum of 3 minutes of rest was given to minimize fatigue (Parcell, 2002). If the participant did not feel adequately recovered, additional rest was allowed. Experimental trials were conducted in a similar fashion to the calibration. The cameras

were started and three claps were performed to facilitate audio synchronization during data processing. The participant was then instructed to perform the set of 5 pushups for that condition. Upon completion of the pushups, the cameras were made to stop recording and the participant was instructed to relax for the given period of rest. This procedure was repeated for all subsequent conditions.

3.8 Data Processing and Statistical Analysis

Raw electromyographic experimental data from each muscle was processed using a root mean square (RMS) method in accordance with previous literature (Farfan, 2010). The same three repetitions that were analyzed for kinematic variance were used for electromyographic analysis. The RMS value was calculated including EMG data from the beginning of the first repetition of study to the end of the third repetition of study. These values were normalized to the maximum absolute value from the MVIC trial, and the resulting %MVIC data were used in statistical calculations to represent relative level of muscle activation for each muscle, during each condition.

Motion capture data was processed using open-source Argus software. Digitization of the raw video files generated x- and y-coordinates for the point of interest. The raw video files from each of the three cameras were first synchronized, then digitized. Digitization of the video files involved manually identifying the center of the point of interest with a mouse click in all thee camera views, for each frame. Calibration video files averaged 600 frames in length (10 seconds at 60 fps), and experimental video files averaged 840 frames in length (14 seconds at 60 fps). Given the three cameras and

two points of interest on the calibration wand (each end of the wand), processing the calibration data required digitization of 3,600 frames per participant. With three cameras recording six points of interest on the LED triads in the experimental video files, across three conditions, processing the experimental data required digitization of an average of 15,120 frames per participant. Given that data was collected from 10 participants, approximately 187,000 frames were digitized.

Upon 2-dimensional digitization of the calibration video files and the experimental video files, the Argus software combined this information to determine relative 3-dimensional coordinates for all digitized points of interest. The 3-dimensional coordinates of the L1 vertebra were then input into an R-code, written by a member of the research team in accordance with previous literature (Graham, 2012a), designed to estimate the kinematic variability of any point of interest through calculation of maximum short-term Lyapunov exponents (Graham, 2012a).

An a priori power analysis was performed (power = 0.80, effect size = 0.25, alpha = 0.05), with a resultant sample size requirement of n = 28. This sample size criteria was not met and results should be considered accordingly. The independent variable for this study was the number of available DOF of the pushup. The dependent variables, modeled separately, were: muscle activity for the external obliques (EO), rectus abdominis (RA), upper erector spinae (ES), and longissimus (LG), as calculated by %MVIC, and the level of dynamic stability of the participant as estimated by short-term maximum Lyapunov exponents (Graham, 2012a). All statistical analyses, including the stability calculations, were performed with R software (R version 3.3.2). Because the dataset was not normal, Friedman's test was used to determine if a statistical difference in muscle activity existed

across conditions for each muscle. Friedman's test was also used to determine if a statistical difference in kinematic variability existed across conditions. Given a significant result, a Wilcoxon signed rank test was used to determine differences between conditions. Alpha values were set to 0.05. No conditioning was used to filter the hand-digitized kinematic data. The number of frames digitized was dependent on the duration of the set of pushups; given a duration of 10 seconds, recorded at 60 frames per second, 600 frames and thus coordinates from each camera would be used for calculations.

3.9 Statistical Hypotheses

H_0^1: There is no difference in 3-dimensional kinematic variance of the L1 vertebra during pushups between conditions with increasing DOF.

H_0^2: There is no difference in muscle activity of the core musculature during pushups between conditions with increasing DOF.

CHAPTER 4

Results

4.1 Kinematic Variance

Friedman's test demonstrated a significant difference in median values (λ_1= 0.0086, SD = 0.14, λ_2 = 0.1148, SD = 0.19, λ_3 = 0.2281, SD = 0.13) of maximum short-term Lyapunov exponents across the different conditions of available DOF, $\chi2(2) = 6.2$, $p < 0.05$. Post hoc analysis demonstrated significant differences between Level 1 and Level 2 ($p < 0.05$), as well as between Level 1 and Level 3 ($p < 0.05$), but no significant difference was detected between Level 2 and Level 3 (Figure 4).

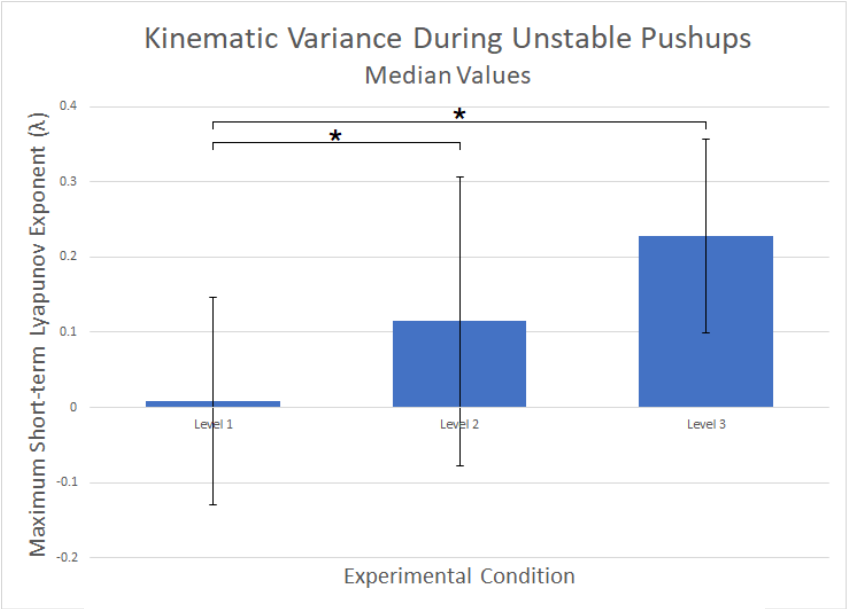

Figure 3: Kinematic variance by level of instability

4.2 Muscle Activity

While muscle activity of the relevant musculature was recorded for all 10 participants, technical problems with EMG data from the first six subjects precluded the use of this data. As such, EMG activity for 4 subjects was used for analysis thanks to an improved system of EMG equipment.

Friedman's test demonstrated a significant difference in median values of anterior muscle (EO, RA) activity across the different conditions of available DOF, save for a near significant result from the left rectus abdominis (Table 1). No significant differences were found among the distributions of EMG recordings from the posterior muscles (ES, LG) of note.

Table 1: Statistical Results - Muscle Activity

		Friedman's	Post-hoc Wilcoxon Signed Rank Test		
			Level 1 <-> 2	1 <-> 3	2 <-> 3
R. EO		**$\chi 2(2) = 12$, $p = 0.0025$**	**$p = 0.031$**	**$p = 0.031$**	**$p = 0.031$**
L. EO		$\chi 2(2) = 4.3$, $p = 0.1146$	--	--	--
R. RA		**$\chi 2(2) = 9.3$, $p = 0.0094$**	**$p = 0.031$**	**$p = 0.031$**	$p = 0.5625$
L. RA		**$\chi 2(2) = 8.3$, $p = 0.0155$**	$p = 0.094$	**$p = 0.031$**	$p = 0.0625$
R. ES		$\chi 2(2) = 0.33$, $p = 0.8465$	--	--	--
L. ES		$\chi 2(2) = 0.33$, $p = 0.8465$	--	--	--
R. LG		$\chi 2(2) = 0.33$, $p = 0.8465$	--	--	--
L. LG		$\chi 2(2) = 1$, $p = 0.6065$	--	--	--

Post-hoc analysis of the right external oblique demonstrated a significant difference across all three conditions of available DOF ($p < 0.05$). Post-hoc analysis of the right rectus abdominis demonstrated a significant difference between Level 1 and Level 2 ($p < 0.01$) and Level 1 and the Level 3 ($p < 0.05$), but no significant difference between the two unstable conditions. Post-hoc analysis of the left rectus abdominis demonstrated a significant difference only between Level 1 and Level 3 only ($p < 0.05$).

CHAPTER 5

Discussion

5.1 Discussion of Results

As the number of available DOF of the pushup were increased, the difference in kinematic variance was statistically significant, with some differences in muscle activity of the anterior core muscles. The differences in kinematic variance were found in comparison to the stable pushup with the fewest available DOF, however no statistically significant differences were found in kinematic variance between each of the unstable pushups. Significant differences in anterior core muscle activity were observed in the right external oblique, and in the rectus abdominis bilaterally. The posterior core muscles saw no significant differences in muscle activity.

5.1.1 Kinematic Variance

Upon analyzing the movement of the L1 marker through 3D space during pushups, short-term kinematic variance is altered as the number of available DOF increases past baseline. In other words, movement of the L1 vertebra appears to change when the environment in which the pushup is performed simply becomes unstable. To our knowledge, the present study is the first to estimate dynamic stability of the human body during exercise in response to an unstable environment. A compromise in dynamic stability of the human body during unstable exercise has been inferred in electromyographic studies (Anderson, 2003; Holtzmann, 2004; Kohler, 2010), and the results of this study suggest that a relationship may exist.

One finding of interest, however, was that kinematic variance does not change in proportion to available DOF, as no significant difference was found between the two unstable conditions of higher available DOF. It seems that kinematic variance may

simply be altered as additional DOF are made available in comparison to Level 1, but may not be altered further. A conceivable explanation of this phenomenon is that as the number of available DOF increases, and the movement becomes more unstable, the participant may be forced to alter their movement mechanics in order to compensate for the unstable environment. This has been suggested in previous studies (Beach 2008; Behm, 2010), but has yet to be confirmed. If the participant were to alter their mechanics by slightly limiting their range of motion, for example, it is plausible that the movement of the marker through 3D space would see a similar level of kinematic variance as a more complete range of motion at a lower number of available DOF. Additionally, the point at which dynamic stability is lost may not be consistent between individuals. Should stability be compromised initially, it is understandable that additional environmental instability may not have quite the same effect.

Another factor which may contribute to this phenomenon is that the progression of instability with available DOF may not be a linear one. It is plausible that adding one available degree of freedom to a baseline stable movement may in fact alter the stability of the movement to a greater degree than progressing through higher numbers of available DOF, such as from five to six. As instability is introduced, the difference in kinematic variance may be more significant than when it is already present and simply progressed. Future research should attempt to examine this phenomenon at higher levels of instability, and could be expanded to observe how a stability training program consisting of movements with many available DOF may affect the maintenance of dynamic stability in conditions with less available DOF.

An important concept to consider is the value of the maximum Lyapunov exponents themselves. For context, previous studies (England, 2007; Graham 2012a; Graham 2012b) calculated maximum short-term Lyapunov exponents within a similar range. It is important to understand, however, that in studies of kinematic variance between different conditions the change in these values may be more valuable than the values themselves.

5.1.2 Muscle Activity

Muscle activity of the anterior core muscles was greater during pushups with greater available DOF when compared to a stable 1-degree of freedom pushup, in accordance to previous studies (Vera-Garcia, 2000; Behm, 2006; Kohler, 2010; Beach, 2008). The degree to which each muscle was activated seemed to increase in relation to the number of available DOF, however statistical analysis indicated the differences were not consistently significant. The right rectus abdominis demonstrated a significant difference between stable and unstable conditions, but no difference between the two unstable conditions, while the left rectus abdominis only demonstrated a significant difference between the stable and most unstable conditions. The left external oblique was not found to exhibit a statistically significant difference in muscle activity, however visual analysis of the data does show a trend of greater muscle activity during the unstable conditions.

While the results indicate that an increased number of available DOF may elicit greater activation of the muscles of the anterior core, it cannot be concluded that this

change in activity occurs at a proportional amount to the number of available DOF. It appears that increasing the number of available DOF of the pushup may be enough to elicit a response of increased core activity, but it cannot be concluded at this time that a concrete relationship exists.

5.2 Considerations

There were several important concepts taken into consideration in the present study in an effort to conduct a valid study. Kinematics and muscle activity have been found to change with hand height during pushups (Cogley, 2005). To control for hand height, the Yoak trainer was suspended at specific heights for each condition to set the level of the hands equal to that of the parallette pushup. While the participants were of different heights and anthropometrics, the alignment of their arms and trunk were held fairly consistent on a relative scale. It should be considered, however, that the conditions of greater available DOF may have been more difficult for longer-armed individuals. While arm length was not measured, it is conceivable that each increase in DOF may challenge an individual with longer arms to a greater degree than it would affect an individual with shorter arms. Width of hand placement has also been found to affect muscle activation patterns in previous studies (Gouvali, 2005). In the present study hand width was controlled for by replicating the self-selected hand width of each participant on the parallettes. After the participant performed a set of pushups of the stable condition, the handles attached to the Yoak were suspended in a manner which would result in a width similar to that of the stable condition.

Practice or experience with suspension pushups would theoretically result in an adaptation of increased efficiency and stability in a participant (Sandrey, 2013). While all participants in this study were resistance trained individuals, no participants indicated any level of experience with regards to instability training. All participants self-reported to have experience with typical training modalities such as barbells, dumbbells, and bodyweight, but not one participant was practiced in unstable environments. This created a level of parity across participants with respect to training level, minimizing the effect that practice or experience on unstable devices may have had on the results.

The original experimental design consisted of pushups with 1 available DOF, 3 available DOF, and 5 available DOF. During pilot testing it was made abundantly evident that it would be extremely difficult to successfully perform the 5-DOF condition, even for highly resistance-trained individuals. Thus, the design was altered to include 1, 3, and 4 available DOF. Even so, upon collecting data, it was evident that the 4-DOF condition was also too difficult for five participants. These participants were thus regressed to a 1 DOF, 2 DOF, 3 DOF protocol. While the specific protocols between participants capable of performing the 4-DOF trial and those incapable of performing the 4-DOF trial were different, the result was nonetheless a protocol of pushups with progressive levels of instability.

Trial randomization is commonly employed to reduce the chance that a systematic bias may influence the results when the same protocol is followed for each participant (Sibbald, 1998). While this may be an effective strategy for some studies, the non-randomized protocol used in this study was out of practical necessity. Consider a participant who was unable to perform the most difficult condition. Should this condition

be presented first, the participant would be at a greater risk of both physical and psychological distress. It was important to determine the skill level of the participant during the less difficult conditions first, to better understand what level of instability may in fact be too much for each specific participant.

5.3 Limitations

There are some limitations which should be taken into consideration in the present study. The statistical power of the presented analysis is very low due to the extremely small sample size; a power analysis calculation suggested a sample of 28 would achieve appropriate statistical power. One important consequence of the small sample size in the present study (n = 10) is the increased risk of Type II error, or failing to reject the null hypothesis when it is in fact false (Faber, 2014). In the case of a false-positive null hypothesis, it is concluded that no statistically significant difference exists, when in fact statistical modeling with a higher power (from a larger sample size) would reveal a significant difference.

Friedman's test was used for analyzing the data as the sample size of the present study was too small to use statistical models of parametric data. While the activity of the anterior core muscles appeared to increase with DOF, and Friedman's test demonstrated a significant difference for three out of the four anterior muscles, further post-hoc analysis of the data resulted in several non-significant results. If these failures to reject the null hypothesis were a product of type II error, it is plausible that a larger study may demonstrate statistical significance in the same comparisons.

The main factor behind the limited sample size was the time requirement for digitization. Collection of one experimental trial typically lasted 50-60 seconds in length. Each trial was recorded on three action cameras, generating 180 seconds of footage recorded at 60 frames per second, or 10,800 frames per experimental trial. Given that each participant performed three trials of different conditions, the result was over 30,000 frames of 2-dimensional video to be processed for every participant. Digitization of the 2-dimensional video segments to be used to generate a 3-dimensional representation of the LED marker's positioning was done on a frame-by-frame basis; the data for each participant took approximately 50 man-hours to digitize and process. Advances in this technology have allowed for this process to become quicker, and would make this portion of the study possible in a much shorter amount of time. With this temporal advantage comes the ability to process and analyze much larger datasets, albeit at an often very expensive price. The methods demonstrated in this study would be better suited to movements which cover a much larger area of interest, such as gross limb movement during a kick or a throw. What this portable and easy-to-use setup permits is the collection of video-graphic human movement data in locations outside of an experimental laboratory, which may result in the observed movement being more specific to the environment in which the participant usually performs the movement.

It is common for studies of human movement to control for movement speed using a metronome or timer (Youdas, 2010; Dyrek, 2011). In the present study, the use of a metronome was considered, but not employed. During pilot testing, it was found to be extremely difficult to maintain a given tempo during execution of pushups with several available DOF. These unstable variations required a high level of neuromuscular control,

and an attempt to match the tempo provided by the metronome compromised the kinematics of the pushup to such a degree that a complete range of motion was never attained. It was determined that a complete movement would be more important to the present study than controlling for speed of movement as stability training typically focuses on quality of movement as opposed to speed of movement.

5.4 Practical Application

Unstable training has been demonstrated in previous literature to be effective for both accelerating rehabilitation programs (O'Sullivan, 1997; Hides, 2001) and eliciting development of the core musculature to a greater degree in populations engaging in resistance training (Vera-Garcia, 2000; Anderson, 2003; Behm, 2005; Beach, 2008). Based on these findings, implementing an environment of instability during exercise may help the participant reach their training goal in a shorter amount of time, saving resources and improving quality of life.

While many products to create this environment of instability have been marketed, currently no established system of quantifying the degree of instability exists. It has been suggested that environments of low to moderate instability do not elicit the same effects in resistance trained individuals as in untrained individuals (Wahl, 2008). For these trained populations, a quantification system needs to be available to ensure the level of instability is sufficient enough to stimulate the core musculature. The present study indicates that the number of available DOF has potential to be a suitable quantifier

of instability; future studies observing stability at a larger range of DOF conditions may help to confirm this theory.

Current methods for analysis of human movement typically require a laboratory environment (Mundermann, 2006); the calibration of the cameras influences the accuracy of the generated 3-D representation (Hedrick 2008), thus a dedicated lab space permits recycling of calibration information to be used in successive experiments. Unfortunately, many of the validated motion capture systems of today are quite expensive, making them less accessible to researchers with limited resources. While accurate, these fixed systems also require the movement to be performed in a laboratory, which may be a foreign environment to the subject, potentially resulting in an unnatural movement (Jackson 2016). The present study proposes a three-dimensional motion capture system which, thanks to its affordability and portability, will facilitate studies of human movement in the natural environment in which the subject normally participates.

5.5 Conclusion

The results of this study suggest that quantifying the available degrees of freedom for potential instability of a movement may be an effective way for healthcare practitioners to control and manipulate the level of difficulty of the movement in question. While not every instance was statistically significant, differences were found between amount of kinematic variance and levels of core muscle activity as the number of available DOF for potential instability increases. Further research of a larger sample size is needed to determine if a relationship does in fact exist. Should a relationship truly

be present, the results would suggest that practitioners could use available DOF as a measure for the amount of instability present in their clients' programs.

5.6 Future Research

While strong conclusions are difficult to make from the results of this study, it does provide a framework for future research upon which to build. In expansion of this study, researchers may want to record a subjective rating of perceived exertion (RPE) from the participants. RPE has been demonstrated in previous literature to be associated with several physiological factors (Pageaux, 2016).

The initial plan of this research study was to determine kinematic variance of the L1 vertebra in relation to the L5 vertebra. Given the preliminary findings of virtually no movement between the two vertebra during the pushup, exercises which involve movement in the transverse plane may be more likely to create this movement. If the goal is to examine movement of the L1 vertebra in relation to the L5 vertebra, unilateral exercises such as the split squat, single arm row, or single leg squat may be more effective selections.

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