THE EFFECTIVENESS OF INSTABILITY RESISTANCE TRAINING DEVICES FOR HIGHLY CONDITIONED INDIVIDUALS

MICHAEL JONATHAN WAHL

The Effectiveness of Instability Resistance Training Devices for Highly Conditioned Individuals

By

©Michael Jonathan Wahl

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A Moment If You Please ...

Moments Can Be Short, Moments Can Be Long

There Are Moments Of Joy, Moments Of Sorrow

Moments Of Passion, Moments You'll Never Forget

Moments You've Already forgotten, Moments You Didn't Get

There Are Awkward Moments, Senior Moments

Moments Of Truth , And Momentary Lapses In Judgment

People Who Ask For A Moment, Share A Moment

I Need A Moment...

You Got A Moment ?

Wait A Moment, You Can Take A Moment

Make A Moment, Spoil A Moment

And if All The Stars Line In The Right Moment...

That Moment Can Be Perfect, Moments Can Define You

Moments Can Delight You, And Moments Can Change Your Life

Here's To The Moment

And Squeezing All You Can Out Of Every Last Single One Of Them

- Glen Hunt

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List of Abbreviations

1 RM –	One Repetition Maximum
APA –	Anticipatory Postural Adjustments
BF –	Bicep Femoris
EMG –	Electromyography
Hz –	Hertz
LAS –	Lower Abdominal Stabilizers
LSES –	Lumbo-Sacral Erector Spinae
Ms –	Millisecond
mV –	Millivolts
MVC –	Maximum Voluntary Contraction
N –	Newton
PF –	Plantar Flexion
RF –	Rectus Femoris
SD –	Standard Deviation
SEC –	Seconds
SBD –	Standing BOSU Down
SBU –	Standing BOSU Up
SD –	Standing Dyna Disc
SS –	Standing Stable
SqBD –	Squatting BOSU Down
SqBU –	Squatting BOSU Up
SqD –	Squatting Dyna Disc
SqSB –	Squatting Swiss Ball
SqS –	Squatting Stable
SqW –	Squatting Wobble
SSB –	Standing Swiss Ball
SW –	Standing Wobble

Co-Authorship Statement

This thesis was prepared under the supervision of Dr. David Behm (School of Human Kinetics and Recreation). Dr. Behm played a vital role in my topic development, proof reading and editing and technical support. I was personally responsible for subject recruitment, data collection, data analysis, and final write up of the enclosed research article “Not All Instability Training Devices Enhance Muscle Activation in Highly Resistance Trained Individuals”.

Chapter 1 Introduction

Background of Study

Stability training is a leading modality in the field of resistance training and athletic conditioning. Its uses have extended beyond their traditional rehabilitative conditions and entered the world of bodybuilding and athletic performance. Several instability devices have emerged as standard fixtures in fitness facilities around the world and are utilized by all populations. Supporters of unstable training can be found in the popular media (i.e. Paul Chek, Juan Carlos Santana) interpreting current instability research and applying it to elite athletes. Interestingly no research, to our knowledge, investigated the effect of these training devices using experienced subjects. Studies have used recreational active, geriatric or rehabilitative subjects to quantify their results.

Previous research indicated increases in electromyographic (EMG) activity when performing exercises under unstable conditions (Marsden et al. 1983, De Luca and Mambrito 1987 Stanforth 1998, Sternlicht, and Rugg, 2003, and, Anderson and Behm 2004a, Behm et al. 2002, 2005). Other studies have found a decrease in force output when performing resistance training under unstable conditions (Behm et al. 2002, Anderson and Behm 2004b). Concurrently some research has found training on unstable devices to show no significant difference in EMG activity (Anderson and Behm 2004b, Behm et al. 2005). However, of the aforementioned studies, no analysis of specific instability training devices were measured using subjects who were highly resistance trained.

Thus, it is important to evaluate the effectiveness of instability resistance training devices when used by individuals who have resistance trained extensively with relatively unstable free weights to accurately assess the EMG response while performing exercises using these tools.

Purpose of Study

Training specificity is a highly investigated aspect of fitness and conditioning. Research into training angle, velocity, and contraction type, which characterize the exercise and modality specificity have formed the basis of exercise prescription in rehabilitation and conditioning (Moffroid, and Whipple, 1970, Behm et al. 1993, McCaw, 1994,). One area which has recently emerged is stability specificity. Instability has been identified as a predictive variable in some sport performances (Behm et al. 2005b) and as such, the quantification of various instability training tools and exercises under unstable conditions is relevant and vital to the development of specific training programs. An objective of this study is to investigate the relative EMG activation of involved trunk and lower body musculature during exercises requiring an unstable base in a population of highly resistance trained individuals. Several studies (Stanforth et al. 1998, Vera-Garcia et al. 2000, Sands and McNeal 2002) have investigated and compared training tools but a gap exists in regard to the EMG activation of trunk and lower body musculature using instability training tools under various conditions including exercises, posture and fatigue. Furthermore, there are no studies to our knowledge that have examined the effectiveness of instability devices in a highly trained group of individuals. Typically, studies using instability devices report significantly greater EMG activation than with stable bases in untrained and recreationally active subjects (Stanforth et al. 1998, Vera-Garcia et al. 2000, Sands and McNeal 2002, Behm et al. 2005a).

The goal of this study is to identify EMG characteristics of involved trunk and lower body musculature during various exercises, postures and fatigue protocols using instability resistance training tools with an experienced group of resistance trained individuals.

Hypotheses

1) It is hypothesized that EMG activity will increase as the base of support becomes increasingly more unstable during both standing and squatting postures.
2) It is also hypothesized that the use of an instability training device (Dyna disc) will demonstrate significant increases in EMG activation during exercise postures versus their stable counterparts.
3) Finally, it is hypothesized that not only will the rate of fatigue be greater under unstable conditions but the EMG activation will be higher in trunk and limb musculature during the fatigue protocol.

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Introduction

Strength training has long been accepted as a means of improving sport performance by increasing either neuromuscular efficiency and/or hypertrophy. Research in training specificity has yielded several criteria and prescription guidelines from which these responses can be elicited. Previous studies have highlighted many aspects of strength training including mode specificity (McCaw 1994, Sternlicht and Rugg 2003), contraction type specificity (Behm & Sale 1993), velocity specificity (Coyle et al. 1981, Behm 1991), and the focus of this review, stability specificity (Stanforth et al. 1998, Vera-Garcia et al. 2000, Anderson and Behm 2004b, Behm et al. 2002, 2005a,b).

Stability is a vital component of sport performance and activities of daily living (ADL). Whether an athlete is jumping off one foot, taking a check in hockey or flipping on the balance beam; possessing optimum stability is an athletic asset. Behm and Wahl et al. (2005) determined that balance was a key predictor in the hockey skating ability of younger players. Behm (1995) states that the more a training movement deviates from the sport movement the less applicable the strength gains will be to the sport. This concept indicates a need to develop sport specific movements and balance challenges in the creation of training regimens.

Stability extends beyond the narrow scope of athletic performance into daily life. Training to increase stability could serve to increase the quality of life of elderly individuals and prevent dangerous falls which can result in serious injury. Several investigations have explored anticipatory postural adjustments in ADL's and more specifically stability in the elderly (Pellec and Malton 2000, 2002, Davis et al. 2004, Gostic 2005). Similarly, workers performing skills under unstable conditions could benefit from a further understanding in regards to improving and determining factors involved in stability. A needs analysis of a sport or workplace should dictate the predominant muscles used, contraction types, range of motion, velocity and metabolism for the training prescription. An often-neglected aspect of training specificity is the balance or stability needs particular to the requirements of daily or optimal function at home, the workplace or the competitive arena.

Training Specificity

Training specificity is a highly investigated aspect of rehabilitation, sport performance training and fitness. Every aspect from speed of contraction, contraction type and modality of training has been dissected to yield a greater training response (Moffroid, and Whipple, 1970, Behm and Sale 1993, McCaw, 1994). Henry (1958) proposed the specificity hypothesis, which states that each person possesses a large number of motor abilities, with each task containing specific abilities that are highly independent of each other, with very low transfer between tasks. Henry (1958) was familiar with studies involving the specificity of intelligence and challenged the idea that a General Motor Ability existed. He therefore proposed the Specificity Hypothesis of Motor Learning, which suggests that the underlying abilities of a motor skill or task are specific to that skill or task and not transferable (task-specific). For example, Bachman (1961) performed an experiment on two tests of balance, and found correlations to be very low, between 0.25, and -.12. Other studies such as those by Lindeberg (1949), Lotter (1960), Parker & Fleishman (1960), and Proteau et al. (1992) support specificity, rather than general motor ability. Parker and Fleishman (1960) utilized factor analysis to examine the abilities which underlie motor skills. The statistical tool can be utilized to correlate several tests to one test. Through this correlative method Parker and Fleishman found sufficiently high correlations to assume that there were indeed general abilities such as coordination. Conversely, according to Schmidt and Lee (1999), even two very similar tasks, such as throwing a football and throwing a javelin will not correlate well with each other. Thorndike and Woodworth (1901) proposed the Identical Elements Theory of Transfer, which suggests that the amount of transfer or benefit, training in one situation would have on another would be determined by the number of elements that the two situations had in common. The following sections identify the elements which must be addressed.

Velocity Specificity

Strength, defined as the amount of force produced externally (Young and Bilby 1993), has long been a concern of scientists, coaches, and athletes. Although maximum force produced is an important variable in sports, very often there are time constraints on which force must be produced. The ability to produce force at a given velocity or speed may be a more relevant means of determining the training regime necessary to improve sport performance. Therefore sports emphasizing strength, speed or power, such as sprinting, throwing and jumping have adopted training protocols, which will develop the maximum amount of force generated over a dictated period of time.

Velocity specificity research states that performance increases occur to their greatest extent at the velocity at which they are practiced (Komi and Tesch 1979, Sale & MacDougall 1981, Sale 1987, 1988, Behm and Sale 1993, Schmidt and Lee 1999). Early evidence of velocity specificity research dates back to the early 1970's, when Moffroid and Whipple (1970) found that subjects who trained at slow speeds (36°/sec) only achieved increases in force production at that training (slow) speed, whereas subjects who trained at a higher speed (108°/sec) experienced increases in all speeds (0 to 180°/sec). Gollnick et al. (1973, 1974) found fast twitch fibers are selectively recruited during high intensity work, while slow twitch fibers are primarily recruited during moderate intensity exercise. Ewing et al. (1990) supports these claims stating that high velocity training will recruit the high threshold motor units and corresponding fast twitch muscle fibers which are selectively stressed during high velocity work, whereas slow twitch fibers are called upon during slower velocity contractions. However, Behm (1991) found evidence against velocity specificity in a study on three groups of subjects performing dynamic contractions with either conventional, hydraulic resistance machine, or with surgical tubing performed at an average angular velocity of 180°/sec. Behm's (1991) experiment found similar increases occurred at all testing velocities on an isokinetic dynamometer, thus indicating an absence of velocity specificity. The lack of velocity specificity training effects were proposed to be due to the range of velocities encountered with the training devices contrasting with the constant angular velocities used to train with isokinetic devices. Behm's results do not deny velocity specificity but serve to demonstrate that a range of velocities may provide a range of adaptations. Defining the training velocity is important for athletic performances where a wide range of velocities might be needed and transfer of specific gains would greatly optimize training (Pereira and Gomes 2003). It has been shown that training should be velocity specific, but related variables such as coordination, angle, contraction type, mode and stability specificity all must be addressed to accurately identify the concepts involved with athletic prescription.

Angle Specificity

Angle specificity states that strength training effects are specific to the criterion joint angle selected during training. Increases in force occur exclusively or are greater at the training angle in comparison to deviated angles (Lindh 1979, Kitai & Sale 1989). An abundance of literature exists confirming this occurrence, particularly in isometric training which has specific relevance to many instability training exercises. Lindh (1979) attributes angle specificity to a neural mechanism, motor skill development, and different recruitment of muscle fibers. Thepaut-Mathieu et al. (1988) support the notion of specific muscle fiber recruitment as they found an increase in EMG activity as well as an improved maximum voluntary contraction (MVC) at the criterion joint angle. Each experimental group was trained during a 5 week period at either 25, 80, or 120° joint angle. Bipolar surface EMG was recorded from the biceps brachii and brachioradialis muscles. MVC increases were always found at the training angle and were systematically greater than at the other angles.

Furthermore, training at shorter muscle lengths resulted in an even smaller range of angle specific gains. An increase of the maximal integrated EMG of both biceps brachii and brachioradialis was correlated with the increased MVC at the training angle. Concurrently, Miaki et al. (1999) agreed reporting angle specific patterns of gastrocnemius integrated EMG activity in their study utilizing isometric contractions in young males throughout different angles of contraction of the triceps surae.

The abundance of isometric contraction research has determined that strength gains are specific to the angle trained (Lindh 1979), but the dominant actions found in athletic training are dynamic contractions. Barak et al. (2004) investigated short range of motions (ROM) with both concentric and eccentric contractions to determine whether angle specifity can be transferred to isometric contractions at different joint angles. Significant increases in isometric force output were found indicating that isokinetic training at limited ROM will increase strength at various angles.

The previous studies have all demonstrated that the changes in length and contraction angle of the muscle during testing can influence the force potential of the muscle due to the force-length relationship. The force length relationship states that; when a skeletal muscle that is actively producing force is shortened or stretched, the resulting steady-state isometric force after the dynamic phase is smaller or greater, respectively, than the purely isometric force obtained at the corresponding final length (Rassier and Herzog 2003). In regards to isometric contractions under unstable conditions one could hypothesize that there is a greater variation in the isometric force output due to the increased activation with unstable conditions versus stable conditions at similar joint angles (Anderson and Behm 2004b). Furthermore, it could be challenged that static contractions performed under unstable conditions are not isometric due to the postural adjustments (Pellec and Maton 2000) associated with an unstable base.

Angle specificity studies have typically used isometric contractions (Lindh 1979) or limited ROM (Barak et al. 2004). However, most sport movements are dynamic and it is important to identify the characteristics of these contraction types in regards to training and performance.

Contraction Specificity

Contraction specificity is a principle which states that optimal increases in strength for a particular contraction occur with practice of that specific contraction and performance increases in one contraction may not necessarily transfer to increases in another (Rutherford et al. 1986). For example, Rutherford et al. (1986) investigated the effects of 12 weeks of isotonic strength training of the quadriceps on isokinetic cycling performance. The results showed a 160-200% increase in the load lifted during training, yet no significant changes in peak power output were generated during isokinetic cycling. In a comparison of isokinetic contractions on isometric performance, Mannion et al. (1992) found 16 weeks of isokinetic training on a leg extension resulted in no significant improvements over a control group on isometric strength gains, yet significant improvements occurred in the experimental group in isokinetic activities (peak pedal velocity and peak power output) while no significant gains occurred in these respective tests for the control group.

However, isometric and dynamic contractions may share similar specific mechanisms for generating peak force. Haff et al. (1997) used eight trained men to compare isometric and dynamic force-time variables, calculated by using percentages of their subjects 1 RM on various levels, 80%, 90% and 100% under both dynamic and isometric contractions. Isometric rate of force development showed moderate to strong correlations with dynamic peak force during 80%, 90%, and 100% 1 RM (r = 0.65, 0.73, and 0.75, respectively) and was strongly correlated with peak dynamic rate of force development during 80%, 90%, and 100% 1 RM (r = 0.84, 0.88, and 0.84, respectively).

This suggests that the ability to exert both isometric and dynamic peak force shares some structural and functional foundation with the ability to generate force rapidly (Haff 1997).

Elaborating on previous research, Haff et al. (2005) investigated isometric versus dynamic peak force development in elite women weightlifters. Again, isometric peak force developments showed moderate to strong relationships to the athlete's competitive snatch, and clean and jerk. However, in this experiment, only 30% of the athletes MVC was used while performing isometric contractions. This supports previous research, which suggested that the intent to lift ballistically elicits the same response as lifting ballistically (Behm 1991).

Contraction and coordination were the focus of several studies in specificity research (Sale and MacDougall 1981, Bobbart and Van Soest 1994, Young 2006). As Young (2006) indicated, specificity of contraction is vital when sports skills are considered. Bobbart and Van Soest (1994) measured intermuscular coordination versus intramuscular coordination in jumping movements. The coordination of the muscles involved in sports movements is defined as intermuscular coordination (Schmidtbleicher 1992), while the internal muscle characteristics and their efficiency such as motor unit recruitment, firing rates, synchronization and reflex potentiation are termed intramuscular coordination (Sale & MacDougall 1981). Bobbart and Van Soest (1994) demonstrated with a computer generated model that force production in the quadriceps, through improved intramuscular coordination, will not improve jump performance but in fact decrease efficiency when intermuscular coordination is impaired. In another study by Morris (2001) quadriceps force was significantly increased during an isokinetic knee flexion/extension (100^{o}) strengthening protocol with no significant effect on long jump performance. Interpreted, this indicates that because jump training was not practiced during the protocol, intermuscular coordination between the involved muscles may not have been optimal irregardless of strength increases. A consideration of the previous research is that untrained subjects have been used who may not have developed ingrained motor programs associated with repetitive skill performance undergone by highly conditioned individuals. Nonetheless, when stability is concerned, one must consider coordination and force production as a delicate balance in order not to disrupt, but to amplify this intermuscular coordination.

The implementation of training plans also follows the concept of training specificity. Although many sports and activities involve a variety of physiological needs (i.e. strength, power, endurance, balance), attempting to train a number of different attributes may lead to training adaptation interference.

Concurrent Training

Physical and neurological adaptations occur specifically as a response to a training stimulus. Athletes contract at different angles, velocities and with different structures in sports which require them to run, jump, throw, and sprint for entire matches. This requires both muscular endurance and strength. To develop these characteristics, trainers employ differing modalities to elicit the desired response; a principle known as specificity of training (Nelson et al. 1990). Entering training camp, it is not unique to train endurance (running), strength (weights), agility, and power in one day, but does this benefit or hinder the athletes?

The relation between concurrent strength and endurance training is referred to as the interference phenomenon (Docherty and Sporer 2000). This phenomenon states that strength gains will be compromised when training simultaneously with aerobic power (Docherty and Sporer 2000). The validity of this model has been challenged due to the lack of controls regarding independent variables common to these studies. Research has yielded an array of conclusions both supporting and denying the validity of the interference phenomenon.

Dudley and Djamil (1985) investigated men and women to identify whether training concurrently for strength and endurance is incompatible. Endurance was trained using a cycle ergometer and knee extension protocols were employed for strength development. Subjects performed three training session per week for seven weeks. The results of the study indicate that concurrent training for strength and endurance does not alter the increase in aerobic power induced by endurance training only. In contrast concurrent training reduced the magnitude of increase in angle specific maximal torque at fast, but not slow, velocities of contraction (Dudley and Djamil 1985). Further research by Nelson et al. (1990) found that simultaneous training of strength and endurance may inhibit the normal adaptation to either training program when performed alone. It was concluded that the extent of the interference depends on the nature and intensity of the individual training (Nelson et al. 1990). Alternately, Sale et al. (1993) investigated the interaction between concurrent strength and endurance training in young men and women. Subjects were divided into two groups where one leg was conditioned using either a strength or endurance protocol. In all subjects, the alternate leg was trained in a combined strength and endurance protocol and all subjects trained three times a week for twenty-two weeks. It was concluded that concurrent strength and endurance training did not interfere with strength or endurance development in comparison to strength or endurance training alone (Sale et al. 1993). Balabinis et al. (2003) agreed, finding that a concurrent strength and endurance regimen does not have any antagonistic effect on either regime. Hennessy and Watson (1994), in an investigation into strength, endurance, power and speed, found that training for strength alone result in gains in strength, power and speed while maintaining endurance. Strength and endurance training, while producing gains in endurance and upper body strength compromises gains in lower body strength and does not improve power or speed (Hennessy and Watson 1994). In typical training regimes entering a competitive season, popular training guidelines (Bompa 1999) prescribe concurrent training for strength and endurance. It is apparent that certain factors must be considered prior to prescription of a training regime targeting specific sport performance characteristics. Not only do these considerations include specificity, and concurrent training but also modality used to derive these attributes.

Mode Specificity

Research into specificity of training has indicated that there are very differing muscular responses to free weight, machine or unstable training, each with their own benefit to athletic performance. For example, free weights allow an unguided lift as opposed to a locked-in motion of lifting seen in machines (McCaw 1994). This is important in understanding the nature of most competitive sports, which require a distinct balance element. The differences between machine and exercise training for athletes have been examined in depth (McCaw 1994, Simpson 1997) but instability devices are relatively new to the world of training. In regards to specificity of training it is important to examine the variables associated with varying modes of training as well as the responses which each elicit specific to the population being trained.

Free Weights vs. Machines

Free weight training has always been an important modality in athletic training. Several studies have identified the advantages of training in free motion lifting patterns (Garhammer 1981, McCaw 1994, Simpson 1997). From these studies we see that free weights have a more applicable transfer to athletic competition and performance compared to machine lifts, which control the ascent and descent of the lifting motion into a locked plane. Free weights allow for flexibility in lifts and a three dimensional array during a lift (McCaw 1994). Sport specific actions can be better mimicked with free weight resistance training, and as such have been widely utilized by strength coaches. As a result, athletes have exhibited increased strength and power using closed chain, full body lifts, which are very similar to the closed chain nature of many sports (Garhammer 1981). These closed chain movements allow, and in some cases require, the lifter to modify velocity and angle of contraction, fully extend and contract joints, and assume postures that machines will not accommodate.

Machine exercises do not allow for the variability of free weights but do extend their importance in the area of rehabilitation. Free weights and some machine exercises allow for a force development throughout the lift as stated by Newton's second law of motion, F=M*A (Garhammer 1981). Therefore free weight dynamic lifts are not isotonic, meaning that there is not a constant contraction through out the range of motion. Contrary to free weights, machine exercises do not allow for such a multi-planar acceleratory aspect due to a defined range of motion. This acceleration is an integral part of sport performance. As well, the deceleration aspect of a free weight lift, requiring stability and muscle lengthening is valuable for strength development (Garhammer 1981).

Sport and activities of daily living (ADL) require the recruitment of stabilizer and prime mover muscles in coordination. Due to its relatively unstable nature, utilizing free weight training could yield a more specific approach as compared to machine training alone. Although there is an abundance of machine resistance exercise equipment in unsupervised or recreation exercise facilities that does not validate its effectiveness as machine training's transferability is negligible to the athletic arena.

Range of Motion with Free Weights

Free weights allow an unguided lift as opposed to a guided motion of lifting used in machines. As stated, locked in range of motion may not be desirable for athletes but also individuals suffering from muscle dysfunction. Comerford and Mottram (2001) identified several variables associated with dysfunction of the muscular system such as myofascial restrictions and connective tissue dysfunctions all of which lead to failure in movement. This implies that if a muscular dysfunction exists in an individual then a locked-in exercise range of motion, which they could not perform correctly due to restrictions, would not aid but in fact amplify this mechanical dysfunction. Furthermore, many machines are performed from a seated position which may not provide enough added resistance to make up for the decreased gravitational pull leading to further deviation from real life movements (Sternlicht and Rugg 2003). Machines are also very often open chained units where muscles are trained from a seated position. This clearly is not similar to the closed chain nature of many sports and as Behm has stated; the more a training movement deviates from the sport movement the less applicable the strength gains will be to the sport. Studies such as Bobbart and Van Soest (1994) and Morris (2001) have shown that strength increases attributed to machine training may not only be negligible but even detriment to muscle activation during athletic movement.

Muscle Activation with Free Weights

Muscle activity differs between exercise modalities even during similar motions due to increased or decreased stabilizer function. McCaw (1994) investigated integrated electromygraphical activity (IEMG) values during the ascent and descent phases of the bench press. IEMG values were calculated and compared between lifts performed with free weights versus a guided weight machine. One-repetition maximums on each mode were calculated at the beginning of the study as a baseline and compared to 60% and 80% 1-RM and IEMG activity of these lifts during the study. The results from this study concluded that there was greater muscle activity during the free-weight bench press, particularly during the lighter 60% 1-RM load. Free weights require control of the bar during the lift therefore enlisting stabilizer muscles to assist in correct form (McCaw 1994). Thus, there can be greater muscle activity when using free weights. Experienced lifters may require less stabilizer recruitment during lifts as they have developed the motor program through practice and learning effect while novices may experience greater stabilizer recruitment due to unfamiliarity to the exercise (McCaw 1994). This stability requirement associated with free weight training using dumbbells and barbells has been further exacerbated through the introduction of resistance training employing instability devices. These tools may increase task complexity, require balance and challenge ingrained motor programs but the effectiveness is dependent upon a spectrum of distinct variables.

The Spectrum of Stability Training

Introduction

Stability is defined as resistance to change, deterioration or displacement (Websters Dictionary 2000). Stability training is a modality of muscular development which encompasses the entire body. It promotes full body coordination through proprioceptive exercise techniques. Several unique pieces of equipment are used such as the Swiss balls, Dyna discs, BOSU balls, and wobble boards. Stability is a very different concept than strength and is not in fact synonymous with strength. Strength assesses muscle function with load while stability assesses motor control regulation of muscle stiffness with no external load (Gibbons and Comerford 2001). This indicates that instability resistance training is the combination of traditional muscle strengthening with proprioceptive conditions involving motor control. The trend towards instability resistance training has sparked studies investigating the biomechanical properties and physiological characteristics of various training stimuli and exercises (Behm 1999, Vera-Garcia et al. 2000, Behm et al. 2002, 2005a,b, Anderson and Behm 2004a,b, Sternlicht and Rugg 2005). These studies have proposed that the advantage of instability devices is increased muscle activation compared to performing similar activities under stable conditions (Behm et al. 2002, 2005a). Practically, balance ability has been shown to be a predictor of hockey skating speed due to the unstable nature of the sport (Behm et al.

2005b). On the other hand, the subjects showed a low correlation between stable force output (1 RM leg press) and skating speed in Div. I Junior (17-20 yr old) hockey players. These results support Bobbart and Van Soest (1994) who stated that intermuscular coordination.in tasks such as jumping or in this case balance and skating transfers to dynamic movements to a greater extent than simply exerting force or increasing activation.

A number of studies have demonstrated increased EMG activity under unstable conditions (Pierce 1998, Vera Garcia et al. 2000, Jeffreys 2002, Anderson and Behm 2004a, Behm et al. 2005a) due to the greater stabilizing responsibilities of the muscles (Anderson and Behm 2004a, Behm et al. 2005a). Behm et al.'s (2005a) inventory of upper body resistance training exercises provides insight into upper body EMG activity under unstable conditions, but like Anderson and Behm (2004b) is limited to the use of a single instability device; the Swiss Ball. The Swiss ball is a common instability tool used in many facets of training ranging from rehabilitative exercise to sport performance training. The effectiveness of this tool has been quantified by several studies (Pierce 1998, Stanforth et al. 1998,Vera Garcia et al. 2000, Jeffreys 2002, Anderson and Behm 2004b, Behm et al. 2005a). While instability exercises have been shown to activate stabilizer muscles (Anderson and Behm 2004b, Behm et al. 2005a), a gap exists in information regarding the quantification of muscle activation using instability devices other than Swiss balls.

Instability Devices

The Swiss ball has become a major training tool in the field of resistance training. It has been shown to elicit significantly higher EMG activity during resistance training exercises at similar sub-maximal intensity (Stanforth et al. 1998, Behm et al. 2005, Clark et al. 2003). Stanforth et al. (1998) investigated the effects of Swiss ball or Resistaball ™ training on abdominal and low back musculature compared to traditional trunk training; in this case an abdominal crunch. In this study a subject pool of 72 women were pre-tested, trained and post-tested in either a traditional or Resistaball™ training group. Stanforth et al. concluded that exercise training with a Resistaball™ was comparable to traditional floor work for training abdominal and back muscles but may be of greater benefit for functional activities that require spinal stabilization (Stanforth et al. 1998). The Resistaball™ group improved significantly more than the traditional group during the post-test of the double leg lowering. This is of significance, as the double leg lowering protocol requires spinal stabilization and utilization of deep stabilizer muscles (Stanforth et al. 1998). Several advantages were pointed out during this study and they were that on the Resistaball™ a greater range of motion was possible, proprioceptive or balance was required and the participant could position themselves in a comfortable position as opposed to a machine (Stanforth et al. 1998). Swiss ball curl-ups also showed an increased EMG in the trunk stabilizers when compared to six traditional abdominal training devices. The increased activation was claimed to be due to its stability component (Clark et al. 2003). This change in EMG activity can attributed to the increased recruitment of muscle fibers responsible for both stabilizing and moving during the exercise motion. This may make the Swiss ball a useful tool for improving sport performance (Clark et al. 2003).

As already shown by Stanforth et al. (1998), the Resistaball^{TM} allows for small, subtle positional changes in subjects position, allowing an overload stress and safely challenging abdominal and back muscles in addition to being able to increase sets and repetitions (Stanforth et al. 1998). From these studies it is apparent that increased trunk stabilizer activation is present under unstable bases (Vera-Garcia et al. 2000). Vera-Garcia et al. (2000) showed abdominal training on an unstable surface increased muscle activity compared to its stable counterpart. A curl–up on a stable surface has been shown to elicit a 21% increase in IEMG activity in the rectus abdominus and a 5% increase in EMG activity in the external oblique, while a curl-up on an unstable surface elicited a 35% increase in IEMG activity for the rectus abdominus and a 10% increase of IEMG activity in the external oblique (Vera-Garcia et al. 2000). Anderson and Behm (2004a) supports these claims in his investigation of stable vs. unstable squats, demonstrating increased activation in the trunk and lower body musculature. These findings correspond to Vera-Garcia who used a curl up as a standard external force (bodyweight) indicating that an increased activation under unstable conditions is necessary to facilitate the force output needed when unstable.

The ability to manipulate body position and angle of contraction is a variable which can be modified allowing similar movement patterns to elicit unpredictable results. Sands and McNeal (2002) investigated a kinematic comparison of four abdominal training devices and a traditional abdominal crunch in an attempt to identify the range of motion of four common training tools for abdominals. Two different pivot devices and two different roller devices were used along with the traditional crunch, and range of motion along four joints was measured during use (Sands and McNeal 2002). The results from this study indicated that all four abdominal devices resulted in less range of motion when compared to a traditional crunch (Sands and McNeal 2002). What these results indicated when compared to the study by Stanforth et al. (1998) is that the Resistaball™ would offer the most range of motion through an abdominal crunch. This is an important factor in training athletes, as they require strength throughout the entire range of motion as is necessary on the field of play.

The research compiled on various instability exercises, and in particular abdominal targeted exercises, is referred to by Dr. Mel Siff in his book Supertraining. According to Siff (2000) abdominal targeted exercises are not effective in developing core strength. His view is similar to what he refers to as the eastern training methodology, stating that exercises such as squats, deadlifts and cleans compound abdominal activation significantly more than abdominal targeted exercises. Hamlyn and Behm (2006) elaborated on the supertraining hypothesis and showed that there is in fact a significant increase in core activation attributed to loading the core during movements such as the squat and deadlift as compared to unstable lower abdominal and back trunk targeted exercises.

Effect of An Unstable Base on Force Output

Due to the increasingly specific prescription of exercise modalities in the field of strength and conditioning, exercises are being quantified for efficiency and applicability, especially those involving instability. Studies such as Kornecki and Zschorlich (1994) reported a considerable loss of muscular force and power exerted on an external object when it became unstable necessitating muscle stabilizing functions in the human motor system. Similarly, Anderson and Behm (2004b) investigated EMG activity and force output during instability training, stating resistance training on an unstable surface may force limb musculature to play a greater role in joint stability at the expense of force production. It was shown that MVC under unstable conditions produced 59.6% less force during chest press than under stable conditions using recreationally active subjects. EMG activity during the press protocols, both stable and unstable, was similar with recreationally active subjects. It was hypothesized that the loss of force without a loss of EMG activity was due to the combination of force output and stabilizing function (Anderson and Behm 2004b). This loss of force with increased activation could be attributed to lack of intermuscular coordination due to unfamiliarity of the task (Young 2006).

Trunk stabilizers are not the only area of stability which has been examined. Behm et al. (2002) investigated the effects of leg extension and plantar flexion under unstable conditions finding that there was a decrease in force output and muscle activation. Although leg extension and plantar flexion exhibited decreased activation and force output, trunk musculature was not examined and may in fact have displayed increased activation (Behm et al. 2002). Contrary to Anderson and Behm (2004b) where lower force output was exhibited but EMG activation remained constant, Behm et al. (2002) reported reduced EMG activation in the recorded muscles. The loss in activation may be attributed to an even greater degree of instability, where higher limb contraction forces would have led to total body balance disruption. Thus in light of these two studies, a particular (moderate) degree of instability can allow similar limb activation levels while greater instability prevents full intensity contraction. Performing exercise on an unstable surface affects several factors including the level of muscle activity and coordination and co-activation of the muscles, which work together in an attempt to stabilize the spine and the whole body (Vera-Garcia et al. 2000).

Trunk Activity

Trunk stabilizers are the foundations from which stability is rooted. Upper and lower extremities produce forces, which are facilitated by trunk stabilizer muscles (Behm et al. 2005a). The importance of trunk stabilizers goes far beyond the scope of daily living and extends into the competitive arena of sports and the rehabilitative region of lower back injury. Exercises taxing this stabilizing muscular system may be beneficial in coping with the forces involved in sport or when considering prevention of low back dysfunction, which is a problem that will affect eight out of ten people in a lifetime (Stanforth et al. 1998). In regard to athletes, many suffer from low back pain despite an active lifestyle. A high success rate is exhibited in non-operative treatment of lumbar pain (96%), when dynamic muscular lumbar stabilization has been trained (Saal 1990). Mechanical low back pain due to hyper-lordotic posturing is frequently seen in young dancers and gymnasts as spondylithesis is three times more common in adolescent female involved in these activities than the general population (Stanforth et al. 1998). Stabilization exercises have been shown to be beneficial for chronic low back dysfunction such as spondylolysis or spondylithesis (Arokoski, 1999). These considerations, although not directly attributed to sport performance, may provide a rationale for employing the use of instability resistance training in athletes who suffer from low back dysfunction. These changes in muscle function associated with training are due to specific alteration in the muscle classed as either physical or neurological adaptations.

Physical Adaptations

Training centers and coaches around the world have developed programs to increase muscle girth in athletes in an attempt to increase sport performance and force production. The question remains as to whether this is the approach necessary to elicit the desired outcome of increased coordination and skill. Muscle force is dictated by a number of factors such as muscle cross sectional area (CSA) and neural efficiency (Antonio 2000). Motor units regulate this neurological process and are comprised of a single motor neuron and the multiple muscle fibers that it innervates (Antonio 2000). Typically stabilizer muscles such as the multifidus and transverse abdominus have a greater ratio of motor neuron to muscle fibers than prime movers due to their proprioceptive nature (Antonio 2000).

Increasing muscle CSA is termed hypertrophy training and has generally been thought of as bodybuilding and rarely associated with stability. Generally speaking, hypertophy is achieved by applying a peripheral stress through movement. As this peripheral stress is increased (i.e. load) then the muscle response generally increases in the form of hypertophy. It is believed that there is a distinct dissociation between hypertrophy and instability. This dissociation may in fact be false. Performed properly, unstable resistance training may elicit the necessary stimulus to achieve optimal gains in factors affecting strength. These factors include muscle CSA, neural adaptation and greater stability through intermuscular coordination (Bobbart and Van Soest 1994), resulting in greater stability and prime mover function. The complexity of exercise movements performed under unstable conditions may be transferable to the dynamic nature of sports, especially games emphasizing a stability component such as ice sports. (Behm et al. 2005b). It is necessary to establish whether exercises performed under unstable conditions exhibit changes in muscle activation of highly trained individuals in order to establish their effectiveness in eliciting adaptation, either physical or neuromuscular.

Neurological Adaptations

Neuromuscular efficiency is defined as the neurological capability of a muscle measured through EMG activity, reflex potentiation, alterations of the co-contraction of antagonist muscles, and altered activation of synergist (Behm 1995). Strength can be affected by a number of neurological factors such as the number of motor units involved in a contraction, the frequency of motor unit firing, synchronization of motor units, and the pattern of motor unit and whole muscle contraction and the degree of neuromuscular inhibition (Stone 2002). Strength training has been shown to improve neurological variables including firing rates, and synchronization under a variety of conditions and contraction types (Desmedt 1977, Antonio 2000).

State of stability is a condition which may also relate to neuromuscular efficiency, which will adapt at different rates according to the complexity of the task or exercise (Antonio 2000). Complex movements such as those requiring stability, including motions of the trunk and lower body (squat, step up) have demonstrated a neuromuscular adaptation phase up to twice as long as simple exercises such as the biceps curl and other non-complex movements (Antonio 2000). Stone (2002) concurs, stating that when properly performed, strength training can induce central nervous system changes causing enhanced motor unit recruitment, an increase in the firing rate, synchronization at lower force inputs, an enhancement of motor unit firing patterns for specific movements and removal of inhibitory influences (Stone 2002). This would increase activation and produce greater force within the muscle (Deschenes et al. 1994). This neuromuscular efficiency may also be synonymous with intermuscular coordination and allow for increased task compliance such as those required in sport (Young 2006). Furthermore, the use of instability resistance training may be a means at eliciting increased strength and coordination in individuals through neuromuscular pathways, as exercises performed under unstable conditions have been shown to elicit greater muscle activation compared to their stable counterparts (Anderson and Behm 2004a, Behm et al. 2005a).

Functional Stability

Functional stability is dependent on integrated local and global muscle function (Comerford and Mottram 2001). Mechanical dysfunctions exist as either segmental (articular) or multi-segmental (myofascial) (Comerford and Mottram 2001). Meanwhile, stability dysfunction is diagnosed by the site and direction of give, or compensation that is related to symptomatic pathology (Comerford and Mottram 2001). The question remains as to whether or not these specific pathologies of the stability system can be trained. As stated earlier, motor unit recruitment can be manipulated and trained, skeletal muscle can develop strength and increase muscle CSA, but can the global muscle system of trunk stability be coordinated? According to (Schmidtbleicher 1992) intermuscular coordination can only be developed by practicing the movement for which coordination is sought, a movement which may not be feasible when dysfunction is present. Pain and pathology are common place in highly trained individuals due to continuous physical exertion. As a consequence the recruitment and motor control of the deep segmental stability system results in poor control of the neutral joint position (Comerford and Mottram 2001). This of course reduces the functional stability of the global muscular system and affects sport performance.

In chronic back pain literature, it has been expressed that local stability muscles exhibit a motor control deficit and that the chronic back pain is not related to strengthening (Gibbons and Comerford 2001). This indicates that it is not strengthening, but learning which the muscle must undergo to function efficiently. The abdominal musculature aids in lumbar stability. When compromised, these stabilizer muscles are direct predictors and causes of low back dysfunction (Clark et al. 2003). The stabilizer muscles are located at the lumbar spine and are composed of the transverse abdominus, the deep segmental fibers of multifidus and the posterior fascicle of psoas major (Gibbons and Comerford 2001). These muscles function as local stabilizers and are the roots of the dysfunction, and must be taught rather than strengthened (Gibbons and Comerford 2001). Core stabilization does not recruit slow motor units only as believed by many (Gibbons and Comerford 2001). Adding resistance also recruits fast motor units,. In addition general strength training should be utilized to increase intrinsic muscle stiffness to support the dysfunction (Gibbons and Comerford 2001).

Several studies have investigated the effect of instability in the deep stabilizer muscles including the lumbo sacral erector spinae, the multifidus, transverse abdominus and related structures (Behm et al. 2002, 2005a, Anderson and Behm 2004a). Research has also compared the activation of prime movers in relation to state of stability in order to identify the global stability system referred to by Gibbons and Commerford (2001). Previous research has shed light on the area of instability but research of this discipline is still a fledgling area. A need exists to examine population specific, muscular activation, in a variety of postures, under unstable conditions using the various devices that create the state of instability.

Summary

In summary, EMG analysis of activities performed on unstable bases has established several criteria which may aid in instability resistance training prescription and transfer to sport performance. Typically, unstable resistance training activities show a similar or greater activation to their stable counterparts but with decreased force output. Although prime movers exhibit less force output and can sometimes exhibit less activation with high levels of instability, it has been shown that trunk stabilizers have an increased activation under these unstable conditions. Stabilizers function to maintain balance and if trainable through unstable exercises may lead to increased sport performance (Behm et al. 2005b). Sale (1988) reported that strength gains are primarily due to neuromuscular adaptation in the initial stages of a training protocol. The coordination of antagonists, agonists, synergists and stabilizers through neuromuscular adaptation may enhance strength gains and allow for an increase in athletic performance (Anderson and Behm 2004a). In practical application, athletic training applying overload to the global and local stability muscles may induce these muscular adaptations leading to increased stability. If an athlete exhibits increased stability, prime movers may elicit a greater force output as their role reverts from a stabilizer function back to a prime mover; in essence increasing force output for sport performance.

Stability training has emerged into the forefront of general population and athletic performance training. Instability devices can be bought at every department store or sports shop. Anyone can enroll in classes using stability balls at their local gym and trainers of all levels promote instability training devices as a means of strengthening core and increasing sport performance (Santana 1999). Research has shown significant evidence that training with an unstable base increases muscle activation of both prime movers and stabilizers as compared to their machine or stable counterparts (Stanforth et al. 1998, Behm et al. 2005a). Studies have correlated balance as being a predictive variable in sport using teenage hockey players (Behm et al. 2004b). This research has created a convincing case for the transfer of stability training to sport performance for some practitioners. However, it is also apparent from previous research, (Bobbart and Van Soest 1994, Morris 2001) that although one variable may be affected by resistance training the transfer to sports may not only be negligible but detrimental. Previous research has indicated that significant differences in EMG activation occur when training under unstable conditions in recreationally trained subjects. The question remains as to whether previous studies could apply to highly trained individuals who often utilize instability training in an attempt to enhance performance.

Thus, a study investigating the effects of postures, specific exercises and fatigue under stable and unstable conditions is necessary to identify the effect of stability specificity on highly trained individuals. Additionally, instability devices providing the unstable training environment must be evaluated in order to establish an accurate prescription of these tools for sport performance.

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Chapter 3

Not All Instability Training Devices Enhance Muscle Activation in Highly Resistance Trained Individuals

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Running Title: Instability-related Muscle Activation in the Highly Trained

Abstract:

The objective of this study was to measure the electromyographic (EMG) activity of the soleus, bicep femoris, rectus femoris, lower abdominals and lumbo-sacral erector spinae (LSES) muscles with a variety of a) instability devices, b) stable and unstable (Dyna Disc) exercises and c) a fatiguing exercise in 16 highly conditioned individuals. The device protocol had participants assume standing and squatting postures while balancing on a variety of unstable platforms (Dyna discs, BOSU ball, wobble board and a Swiss ball) and a stable floor. The exercise protocol had subjects performing, static front lunges, static side lunges, one leg hip extensions, one leg reaches, and calf raises on a floor or an unstable Dyna disc. For the fatigue experiment, a wall sit position was undertaken under stable and unstable (BOSU ball) conditions. Results for the device experiment demonstrated increased EMG activity for all muscles when standing on a Swiss ball and all muscles other then the rectus femoris when standing on a wobble board. Only lower abdominals and soleus EMG activity increased while squatting on a Swiss ball and wobble board. Devices such as Dyna discs and BOSU balls, did not exhibit significant differences in EMG activation under any conditions, except the LSES in the standing Dyna disc conditions. During the exercise protocol there were no significant changes in EMG activity between stable and unstable (Dyna disc) conditions. With the fatigue protocol, soleus EMG activity was 51% greater with a stable base. These results indicate that the use of moderately unstable training devices (i.e. Dyna disc, BOSU ball) do not provide sufficient challenges to the neuromuscular system in highly resistance trained individuals.

Keywords: stability, balance, electromyography, strength, fatigue.

Introduction

The importance of optimal balance and stability for athletes is essential for performance and injury prevention (Comerford and Mottram 2001). Instability devices are common in fitness facilities as a means of training. There is an abundance of training methodologies and exercises implementing various instability training devices. The popular media and practitioners (Paul Chek, Juan Carlos Santana) endorse and sell these products promoting unstable training as a means of improving sport performance, force production and core strength (Santana 1999).

A number of studies from the Memorial University Human Kinetics laboratory have reported increased muscle electromyography (EMG) activity when an exercise was performed with an unstable rather than a stable base (Anderson and Behm 2004a, Behm et al. 2005). Anderson and Behm (2004a) demonstrated an increase in activation of the lower body and trunk musculature when performing squats under unstable conditions using Dyna discs. Similarly, Behm et al. (2005a) demonstrated increased trunk activation when performing upper body unilateral and bilateral contractions on a Swiss ball. All of the aforementioned studies as well as other similar studies (Pierce 1998, Stanforth et al. 1998, Vera Garcia et al. 2000, Jeffreys 2002) have used sedentary, elderly or recreationally active individuals. There are no studies to our knowledge that have evaluated instability training with individuals who have trained extensively with relatively unstable free weights. Furthermore, no studies of which we are aware, have compared the impact of a wide variety of instability devices (BOSU Ball, Dyna Disc, Swiss Ball, and Wobble Board) on EMG activity of the lower body and trunk musculature. In this study, a variety of devices, postures, exercises and fatigue conditions were compared to investigate whether unstable conditions and the devices which create these conditions are beneficial to highly resistance trained individuals.

Hypotheses

Three distinct protocols were implemented. The objective of the first protocol was to determine differences in EMG activity while standing and squatting on a variety of unstable platforms and a stable floor. The objective of the second protocol was to examine the EMG activity associated with a variety of exercises that were performed under stable and unstable (Dyna disc) conditions. Since significant differences in muscle EMG activity have been reported between instability devices, exercises, and base of support (Anderson and Behm 2004a, Behm et al. 2005a, Clark et al. 2003, Stanforth et al. 1998,), it was hypothesized that the EMG activity would increase as the base of support became increasingly more unstable.

The objective of the third experiment was to investigate the extent of EMG activity associated with a fatigue test performed under stable and unstable (BOSU ball) conditions. Since previous instability research has demonstrated decreased force production with similar (Anderson and Behm 2004a) or decreased activation (Behm et al. 2002), it was hypothesized that an unstable base would contribute to an earlier onset of fatigue and higher EMG activity.

Materials and Methods:

Subjects

Sixteen subjects (Table 3.1) participated in the study (26.6 ±7.0 yrs, 81.8 ± 9.1 kg, 176.7 ± 8.0 cm). All subjects were considered highly experienced resistance trainers due to their previous and current resistance-training experience (8.2 ± 7.4 yrs) and their extensive involvement with resistance training activities involving free weights and instability devices. Subjects' upper and lower body strength ratio's were determined using the American College of Sports Medicine's (ACSM) guidelines for exercise testing and prescription (ACSM 2000). Mean upper and lower body strength ratio were 1.49 ± 0.17 and 3.18 ± 0.37 respectively. Both mean values exceed the 90 percentile of the male population for strength indicating subjects were considered well above average (ACSM 2000), or for this study; highly conditioned (Table 3.1). Each subject was required to read and sign a consent form before participation. The Human Investigation Committee, Memorial University of Newfoundland, approved this study.

Table 3.1: Subjects' anthropometric measures, 1 RM Bench Press, 1 RM Leg Press, and Upper and Lower Body Strength Ratios

subject	age	Height (cm)	Weight (kg)	years training	1 RM bench (kg)	1 RM leg press (kg)	upper body ratio	lower body ratio
1	23	180	79.5	6	136	273	1.72	3.43
2	24	175	73	2	109	227	1.49	3.11
3	23	180	80	5	136	273	1.70	3.41
4	35	182.5	91	20	125	277	1.37	3.05
5	30	165	70.5	9	125	277	1.77	3.93
6	22	175	81.8	2	110	205	1.34	2.50
7	23	170	72.7	4	108	227	1.49	3.13
8	21	167.5	75	6	105	250	1.39	3.33
9	24	170	84.1	5	148	295	1.76	3.51
10	49	180	97	30	136	295	1.41	3.05
11	28	167.5	72.7	14	114	250	1.56	3.44
12	24	197.5	93.2	8	107	239	1.15	2.56
13	23	180	100	5	136	289	1.36	2.89
14	25	175	79.5	5	109	280	1.37	3.52
15	23	180	80	5	118	250	1.48	3.13
16	29	182	78	5	118	227	1.52	2.91

Mean Values:

AGE:	26.6 ± 7.0
WEIGHT:	176.7 ± 8.0
HEIGHT:	81.8 ± 9.1
YEAR TRAINING:	8.2 ± 7.4
1 RM BENCH PRESS (kg):	121.3 ± 13.7
1 RM LEG PRESS (kg):	258.4 ± 27.9
UPPER BODY RATIO:	1.49 ± 0.17*
LOWER BODY RATIO:	3.18 ± 0.37**

*Upper body strength ratio's >1.48 = 90th percentile or "well above average for 20-29 yr old males

**Lower body strength ratio's > 2.27 = 90th percentile or "well above average for 20-29 yr old males

†1 RM's and strength ratio's were determined using ACSM's guidelines for exercise testing and prescription (weight pushed/body weight)

†Data provided by the institute for aerobics research, Dallas, TX (1994). Adapted from ACSM's Guidelines for Exercise Testing and Prescription. 5th ed. Study population for the data set was predominately white and college educated.

Experimental Design

After an orientation session involving 2-3 repetitions of all stable and unstable exercises on a separate day, subjects performed activities involving both stable and unstable exercises over two separate sessions. EMG activity was recorded during each session. Surface electrodes were placed on the rectus femoris, soleus, and biceps femoris, the lower abdominals and the lumbo-sacral erector spinae (LSES) in order to record the EMG over a 5s duration. For the first (device) protocol, the subjects were randomly assigned to both standing and squatting postures using the following devices: a) Swiss Ball, b) BOSU Ball, c) Dyna Disc, and d) wobble board. Subjects were also randomly assigned to one of the fatigue protocols (stable versus unstable) during the initial session.

The second protocol was completed within 48 hrs. This session required subjects to perform various lower body stable and unstable isometric holds of the following exercises: a) front lunge b) side lunge c) hip extension d) a reach e) calf raise. Exercises performed under unstable conditions used a Dyna Disc under the foot of the load bearing limb.

Finally, subjects performed the remaining fatigue protocol; an isometric wall sit to failure under stable and unstable (BOSU Ball) circumstances. Subjects were instructed to sit against a wall, un-supported, and maintain a knee angle of 90^{o} with their feet either on the floor or a BOSU Ball (concave surface down) until muscular failure occurred (Table 3.2).

Table 3.2: Experimental Design

Objective: to investigate muscle activation changes associated with instability training devices, exercises and fatigue.				
Protocol	Unstable Base	Stable Base	Exercises	Analysis
A) Devices	1-Swiss ball 2-Dyna disc 3-BOSU up 4-BOSU down 5-Wobble board	1-Floor	1-Standing 2-Squat	Separate 1 way repeated measures ANOVAs for stand and squat
B) Exercises	1-Dyna Disc	1-Floor	1-Static Forward Lunge 2-Static Side Lunge 3-One Leg Hip extension 4-One Leg Reach 5-Calf Raises	2 way repeated measures ANOVA Exercise x Base (5 X 2)
C) Fatigue	1-BOSU up	1-Floor	1-Wall Sit 1st contraction, contractions at 1/3 and 2/3 of fatigue duration and final contraction	2 way repeated measures ANOVA Base x Time (2 x 4)
In all three experiments, the following muscles were monitored and analyzed: Biceps femoris, rectus femoris, soleus, lower abdominals, lumbo-sacral erector spinae				

Measurement and Instrumentation

Dependent Variables:

Electromyography: Bipolar surface EMG electrodes were used to measure signals from the LSES, lower abdominals, biceps femoris, rectus femoris and soleus. General descriptive (i.e. LSES, lower abdominals,) rather than specific (i.e. multifidus, longissimus, transverse abdominus, internal obliques) trunk muscle terminology was used in this paper based on the conflicting findings of similar studies. A number of studies have used a similar L5-S1 electrode placement (2 cm lateral to the L5-S1 spinous process) to measure the EMG activity of multifidus (Danneels et al., 2001; Hermann and Barnes, 2001; Hodges and Richardson 1996; Ng et. al. 1998). In contrast, Stokes et al. (2003) reported that accurate measurement of the multifidus requires intramuscular electrodes. Thus, the EMG activity detected by these electrodes in the present study is referred to as LSES muscle activity. Erector spinae muscles according to anatomic nomenclature include both superficial (spinalis, longissimus, iliocostalis and deep (multifidus) vertebral muscle (Jonsson, 1969; Martini, 2001). Additional electrodes were placed superior to the inguinal ligament and 1 cm medial to the anterior superior iliac spine (ASIS) for the lower abdominals. McGill et al. (1996) reported that surface electrodes adequately represent the EMG amplitude of the deep abdominal muscle within a 15% root mean square (RMS) difference. However, Ng et al. (1998) indicated that electrodes placed medial to the ASIS would receive competing signals from the external obliques and transverse abdominus with the internal obliques. Based on these findings, the EMG signals obtained from this abdominal location are described in the present study as the lower abdominals, which would be assumed to include EMG information from both the transverse abdominus and internal oblique.

All electrodes were placed on the right side of the body. Skin surfaces for electrode placement were shaved, abraded, and cleansed with alcohol to improve the conductivity of the EMG signal. Electrodes (Kendall Medi-trace 100 series, Chikopee, MA) were placed on the soleus, 2cm distal to the gastonemius head, as well as the mid-belly of the rectus femoris and biceps femoris. Electrode placement was identified for the second testing session by using indelible markers on the electrode sites for the first session. The EMG signals were amplified (Biopac Systems MEC 100 amplifier, Santa Barbara, CA), monitored, and directed through an analog-digital converter (Biopac MP100) to be stored on the computer (Sona, St. John's, Newfoundland, Canada). The EMG signals were collected over 15s at 2000 Hz and amplified (500X). EMG activity was sampled at 2000 Hz with a Blackman -61dB band-pass filter between 10-500 Hz, amplified (Biopac Systems MEC bi-polar differential 100 amplifier, Santa Barbara CA., input impedance = 2M, common mode rejection ratio > 110 dB min (50/60Hz), noise >5 UV), and analog-to-digitally converted (12 bit) and stored on personal computer for further analysis.

The integrated EMG activity was calculated over a 5s interval of the 15s data collection period; started by the investigator once the subject was in position. Calculations began 5s following the start of data collection and ceased 5s prior to the finish of the specific exercise. The initial and final 5s of collected EMG activity were discarded to minimize postural adjustments at the start or possible fatigue at the end. There was no need to normalize the signal to a maximal voluntary contraction (MVC) since the experiment was a repeated measures design comparing within individuals with all conditions performed within 2 days and electrode placement precisely outlined by a marker. The stable condition was considered as the reference condition to which all unstable EMG activity was compared.

Following the instability device and exercise testing, a fatigue test was performed. EMG changes were monitored by comparing the initial contraction, the contraction at the initial third of fatigue duration, contraction at the second third of fatigue duration, and final contraction. Integrated EMG was collected for 15s of each time period monitored during the fatigue test. The first 5s of each monitored epoch was recorded and analyzed.

Independent Variables

(i) Instability Devices:

All subjects attended an orientation session at least 24 hours prior to testing to familiarize themselves with the exercises.

During the initial testing, exercises were performed with a random allocation technique on the floor (Fig 3.1), a Swiss ball (55cm) (Fig 3.2), Dyna disc (30cm) (Fig 3.3), BOSU ball (55cm) concave surface up (Fig 3.4), BOSU Ball (55cm) concave surface down (Fig 3.5) and a 40 cm wobble board (Fig 3.6). The subjects were positioned either in a standing posture or in a squatting posture with 60° flexion at the knee (measured prior to and during testing using a goniometer). Both postures placed the feet 30 cm apart. Exercise postures were held for a 15s period started by the investigator once the subject was in correct position. All exercises were performed during a single experimental session with a 2 minute rest between each exercise. The exercises were performed twice. Exercises included both stands and squats on both stable and unstable using the instability devices.

(ii) Instability Exercises:

The second session identified the extent of activation with a variety of stable and unstable lower body exercises. Dyna discs were used to create an unstable base for the tested leg in an attempt to identify activation using EMG. The subjects were positioned on the Dyna disc to ensure the orientation of trunk musculature and angle of hips and knees were similar to their stable counterparts.

Static Forward Lunge: Assuming a long lunge position, the participants positioned their back knee 1 cm above the floor while keeping the front knee (90^{o}; measured prior to and during testing using a goniometer) over the ankle. Subjects were instructed to keep the head and chest up and position the hands behind their head to maintain back posture while lowering their hips. The knee of the back leg was slightly flexed (Fig. 3.7). For unstable testing, the dyna disc was placed under the mid-foot of the front leg (right foot). (Fig 3.8)

Static Side Lunge: Subjects were instructed to stand with feet roughly 1.2 m apart and were told to sit to their right side keeping the weight on the right heel as they sat to a 75^{o} (measured prior to and during testing using a goniometer) knee angle. Subjects were instructed to keep the head and chest up and position the hands behind their head to maintain back posture (Fig 3.9). For unstable testing, the dyna disc was placed under the mid-foot of the bent leg (right foot) (Fig 3.10).

One Leg Hip Extension: Subjects were instructed to lie supine with their left leg extended towards the ceiling at 90^{o} (measured prior to and during testing using a goniometer) from the floor. The right foot was placed flat on the floor or dyna disc. The subjects were then instructed to lift their hips while evenly distributing the force over their foot holding this position for 15s (Fig 3.11). For unstable testing, the dyna disc was placed under the mid-foot of the active leg (right foot). (Fig 3.12)

One Leg Reach: Subjects were instructed to stand with their right foot on the floor (Fig 3.13) or dyna disc (Fig 3.14), then reach with their left hand and touch a point 20cm from the front of their right foot. Subjects were instructed to bend at both knees to maintain balance and to achieve both hip and knee flexion.

Calf Raises: Subjects were instructed to balance on their right foot either on the floor (Fig 3.15) or the dyna disc without holding onto any supports. They were then told to plantar flex until fully extended. (Fig 3.16)

(iii) Fatigue

The third protocol identified the rate of fatigue while performing a wall sit under stable (Fig 3.17) and unstable conditions (Fig 3.18). A BOSU ball was used to create an unstable base. Subjects assumed a sitting position against a wall with a knee angle of 90° (measured prior to testing using a goniometer) as well as a hip angle of 90° and feet spaced 30 cm apart. For unstable tests, subjects placed their feet 30 cm apart on the flat side of the BOSU ball (convex surface on floor). The testing was completed when subjects could no longer hold the specified exercise posture. Subjects were instructed to relax when visual inspection indicated a significant deviation of 5° or more from the initial 90° knee angle.

Subjects were analyzed by comparing the rate of fatigue under each condition (stable and unstable) using time as well as an EMG comparison during the protocol.

Statistical Analysis

In the initial investigation (standing and squatting on a variety of instability devices), statistics were performed separately on each muscle group. Data were analyzed with separate 1-way analyses of variance (ANOVA's) with repeated measures for standing and squatting respectively. The six platforms to be compared were the Swiss ball, Dyna disc, BOSU ball up and BOSU ball down, wobble board, and stable floor.

In the second investigation (a variety of exercises performed on a Dyna disc and the floor), data were analyzed with a 2-way ANOVA with repeated measures on the levels. The 2 levels (2 X 5) were state of stability (stable or unstable) and exercise (front lunge, side lunge, hip extension, reach, calf raise).

The fatigue investigation utilized a 2 way ANOVA repeated measures (2 X 4) to determine if significant differences occurred with the EMG activity between the stability condition and fatigue duration (first contraction, contraction at first third of fatigue duration, contraction at two thirds of fatigue duration and final contraction). A 1 way ANOVA repeated measures were used to distinguish significant differences in fatigue duration between stable and unstable conditions.

For all protocols, where significant differences were detected ($p<0.05$), a Bonferroni (Dunn) procedure was used to identify the individual differences among the exercises. Effect sizes (ES) were reported in parenthesis within the results (Rhea 2004). Reliability was assessed with a Cronbach model intra-class correlation coefficient (ICC) (McCaw and Wong 1996) with all subjects (Table 3.5). Repeated tests were conducted within a single testing session.

Results

Instability Devices: Standing

The 1 way ANOVA repeated measures for the device protocol indicated that wobble board had when had 51%, 44%, 43%, and 38% greater soleus EMG activity than standing, the stable floor, Dyna disc, BOSU down and BOSU up ($p<0.004$, ES=0.65, 0.57, 0.56, 0.49). Concurrently, Swiss ball showed 34%, 26%, 24%, and 17% greater soleus EMG activity than table standing, Dyna disc, BOSU down and BOSU ($p<0.004$, ES=0.41, 0.30, 0.28, 0.20)(Table 3.3).

The lower abdominals, during wobble board had 34%, 26%, 33%, and 33% greater EMG activity than stable standing, Dyna disc, BOSU down and BOSU up conditions ($p=0.03$, ES=0.48, 0.36, 0.46, 0.49). Similarly, standing Swiss ball had 31%, 22%, 30%, and 32% greater lower abdominals EMG activity than stable standing, Dyna disc, BOSU down and BOSU up conditions ($p=0.03$, ES=0.46, 0.33, 0.45, 0.48)(Table 3.3).

When standing, the rectus femoris, during Swiss ball had 88%, 61%, 64% and 64% respectively, greater EMG activity than stable, Dyna disc, BOSU down and BOSU up ($p<0.0001$, ES=1.08, 0.77, 0.41, 0.80)(Table 3.3).

The biceps femoris (BF), during wobble board standing had 70%, 65%, 56% and 53% less EMG activity than stable standing, Dyna disc, BOSU down and BOSU up conditions wobble board ($p<0.0001$, ES=1.21, 1.13, 0.98, 0.95). Correspondingly, during Swiss ball had and 57%, 49%, 36% and 33% greater BF activity than stable standing, Dyna disc, BOSU down and BOSU up conditions ($p<0.0001$, ES=1.21, 1.06, 0.81, 0.75)(Table 3.3).

During standing postures, the LSES showed 52% greater activity during Dyna disc than stable conditions ($p<0.0001$, ES=0.73). The LSES also had 68%, 44%, and 42% less activity in stable, BOSU up, and BOSU down conditions as compared to the wobble board ($p<0.0001$, ES=1.36, 0.89, 0.84). In tandem, LSES exhibited 70%, 47% and 46% less EMG activity during Swiss ball conditions as compared to the stable, BOSU up and Bosu down ($p<0.0001$, ES=1.97, 1,32, 1.26) (Table 3.3). There were no significant differences in LSES EMG activity between the Dyna disc and the wobble board or Swiss ball respectively.

Table 3.3: Mean values for iEMG (mV.s) for the standing posture.
Asterisks (*) indicate a significant difference from other unmarked (no asterisks) values in the row. The omega ($^{\Theta}$) symbol denotes significant differences between the two values designated by omegas. The LSES activity with the Dyna disc was not significantly different from the wobble board or Swiss ball.

Muscle	Stable	Dyna disc	BOSU Up	BOSU down	Wobble Board	Swiss Ball
LSES	0.10^{Θ}±0.06	0.21^{Θ}±0.15	0.18±0.11	0.18±0.09	0.32*±0.15	0.33*±0.11
LOWER ABDOMINALS	0.12±0.07	0.13±0.09	0.12±0.08	0.12±0.07	0.18*±0.12	0.17*±0.11
RECTUS FEMORIS	0.02±0.03	0.08±0.06	0.06±0.05	0.07±0.05	0.12±0.10	0.19*±0.15
BICEPS FEMORIS	0.04±0.03	0.05±0.04	0.06±0,04	0.07±0.04	0.14*±0.09	0.10*±0.05
SOLEUS	0.28±0.45	0.32±0.45	0.33±0.45	0.36±0.45	0.58*±0.50	0.43*±0.35

Instability Devices: Squats

During squatting, the wobble board showed 69%, 43%, 57%, 49% more activity in the soleus than stable, Dyna disc, BOSU up, BOSU down conditions respectively ($p<0.0001$, ES=0.91, 0.58, 0.76, 0.64). As well, the soleus, during Swiss ball conditions, showed 54%, 18%, 40%, and 25% more activity than stable, Dyna disc, BOSU up and BOSU down conditions ($p<0.0001$, ES=0.77, 0.24, 0.53, 0.35) The lower abdominals showed 39%, 57%, 48% and 63% (p=0.0002, ES=0.49, 0.64, 0. 71, 0.54) more activity with the wobble board than stable, Dyna disc, BOSU down and BOSU up. Likewise, the Swiss ball exhibited 38%, 56%, 47% and 62% more activity in the lower abdominals than the stable, Dyna disc, BOSU down, and BOSU up conditions (p=0.0002, ES=0.58, 0.79, 0.86, 0.67)(Table 3.4). There were no significant differences among the conditions for LSES, rectus femoris or biceps femoris EMG activity.

Table 3.4: Mean values for iEMG (mV.s) for the squatting posture.
Asterisks (*) indicate significant difference from other unmarked (no asterisks) values in the row.

Muscle	Stable	Dyna disc	BOSU Up	BOSU down	Wobble Board	Swiss Ball
LSES	0.76±0.90	0.52±0.34	0.46±0.25	0.68±0.82	0.80±0.82	0.42±0.22
LOWER ABDOMINALS	0.11±0.08	0.08±0.09	0.07±0.06	0.09±0.07	0.18*±0.15	0.18*±0.14
RECTUS FEMORIS	0.48±0.31	0.55±0.50	0.51±0.50	0.49±0.44	0.42±0.30	0.50±0.32
BICEPS FEMORIS	0.05±0.02	0.06±0.07	0.07±0.05	0.08±0.05	0.10±0.08	0.07±0.03
SOLEUS	0.13±0.08	0.23±0.14	0.17±0.11	0.21±0.13	0.41*±0.30	0.28*±0.19

Instability Exercises

There were no significant differences detected between any of the exercises performed on a stable floor and the unstable Dyna disc.

Fatigue-related EMG Activity

Main effects were discovered for instability conditions and time during the fatigue testing with the soleus. With data collapsed over time, the stable soleus had 51.2% greater activity than the unstable soleus (p=0.03, ES=1.01). There were no other significant muscle activity differences between stable and unstable conditions. Overall, with data collapsed over instability conditions, the last contraction had 36.1% greater soleus EMG activity than the first contraction (p=0.0008, ES=0.33). The interactions illustrated that under stable conditions, the last contraction had 46.4% and 34.5% more soleus EMG activity than the first and second contractions respectively (p=0.001, ES=0.39, 0.84).

Similarly, with data collapsed over instability conditions, the lower abdominals exhibited a 44% increase in EMG activity during the final contraction as compared to the second contraction (p=0.003 ES=0.50). The biceps femoris also exhibited a 35% increase in activity during the final contraction compared to the first contraction (p=0.001, ES=0.55).

As for the fatigue time there was only a trend (p=0.09, ES=1.1) for longer wall sit times under stable conditions (Fig 3.19).

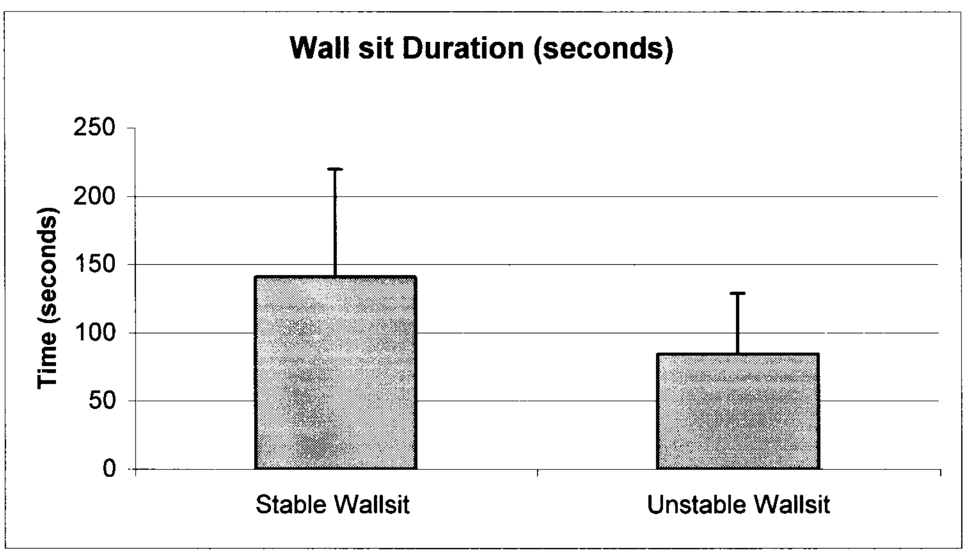


Figure 3.19. The graph depicts the mean time in seconds of the stable and unstable wallsit exercise. . Vertical bars represent SD.

Reliability

Intraclass correlation coefficients illustrated the very good to excellent reliability of the procedures (Table 3.5).

Table 3.5 – ICC Reliability

Reliability was assessed with a Cronbach model intra-class correlation coefficient (ICC) with all subjects. Repeated tests were conducted within a single testing session. Acronyms represent the following: SS: stable standing, SW: standing wobble board, SD: standing Dyna disc, SBD: standing BOSU ball down, SBU: standing BOSU ball down, SSB: standing Swiss ball, SqS: Stable squat, SqW: squat on wobble board, SqD: squat on Dyna disc, SqBD: squat on BOSU ball down, SqBU: squat on BOSU ball up, SqSB: squat on Swiss ball.

	SS	SW	SD	SBD	SBU	SSB
Soleus	0.96	0.9	0.92	0.89	0.98	1
Biceps femoris	0.91	0.9	0.98	0.9	0.92	0.96
Rectus femoris	0.73	0.86	0.98	0.8	0.82	0.82
LSES	0.94	0.97	1	0.97	0.96	1
LAS	0.95	0.9	0.99	0.72	0.96	0.99
	SqS	SqW	SqD	SqBD	SqBU	SqSB
Soleus	0.95	0.9	0.99	0.73	0.96	0.99
Biceps Femoris	0.87	0.9	0.9	0.89	0.97	0.72
Rectus Femoris	0.96	0.96	0.98	0.93	0.96	0.93
LSES	0.94	0.89	0.97	0.72	0.94	0.73
LAS	1	0.97	0.97	0.99	0.99	0.98

Discussion

The most unique finding of this study was the lack of increase in muscle activation (EMG) of experienced resistance trained individuals with activities performed on the unstable bases provided by Dyna disc and BOSU balls. This finding applied in the first (device) protocol to the soleus, rectus femoris, biceps femoris, and lower abdominals when standing on a Dyna disc or a BOSU ball. It applied to the all muscles tested when squatting on a Dyna disc and BOSU ball. It also applied to all muscles tested in the second (exercises) protocol for the exercises performed on a Dyna Disc. Finally, the lack of activation differences for the rectus femoris, biceps femoris, LSES and lower abdominals were also applicable to the wall sit fatigue test performed on a BOSU ball. This is the first study to use experienced resistance trained individuals to demonstrate a lack of significant difference in muscle activation when comparing moderately unstable balance devices to a stable base. Similar to previously published research, the apparently greater instability of the Swiss ball and wobble board did result in greater muscle activation than found with a stable surface and specific to this study, generally greater muscle activation than Dyna discs and BOSU balls (Table 3.6 & 3.7).

Current research both complements and challenges the findings of this study. Several studies have investigated the neuromuscular responses to training under stable and unstable bases using different exercises, tools and populations (Stanforth 1998, De Luca and Mambrito 1987 and Marsden et al 1983., Anderson and Behm 2004a,b, 2005). Cosio-Lima et al. (2003) showed a significant increase in trunk muscle EMG activity and balance scores with unstable versus stable trunk training program in previously untrained women. Vera-Garcia et al. (2000) demonstrated that a curl–up performed under unstable conditions significantly increased rectus abdominus and external oblique activation over curl-ups performed on a stable base. Behm et al. (2005) found similar results, indicating that unilateral upper body exercises as well as lower abdominals and LSES targeted callisthenic exercises, performed under unstable conditions, exhibited greater EMG activity than their stable counterparts. Anderson and Behm (2004a) reported higher soleus and LSES EMG activity when squats were performed on a Dyna disc versus a stable floor. Interestingly the present study did not show any significant difference in activation between stable and moderately unstable (Dyna discs and BOSU balls) exercises. However, the aforementioned studies did not use experienced resistance trained individuals whose balance may have been augmented by years of training (mean 8.2 years ± 7.4) with free weights.

According to Schmidt and Lee (1999), even two very similar tasks, such as throwing a football and throwing a javelin will correlate nearly zero with each other. Conversely, our study found very similar EMG values between exercises performed under moderately unstable (Dyna disc and BOSU ball) and stable conditions. It could be speculated; resistance training with free weights provides an environment of low to moderate instability where learned motor programs may be transferred to other moderately unstable platforms. In accordance with the concept of training specificity, training with moderately unstable free weights transfers to other similar moderately unstable exercises. This may indicate why no significant differences were shown between two apparently different environments. De Luca (1987) showed that EMG decreased with the uncertainty of movement and increased with task awareness. Highly resistance trained individuals who have performed years of resistance training with moderately unstable free weights have become accustomed to specific exercises and therefore have created a strong familiarity with the movements resulting in augmented EMG activity. This experience could reduce the unpredictability of an exercise performed on a moderately unstable tool (Dyna discs and BOSU balls) due to the ingrained motor program of the exercise. Regardless of cause, the current study shows that not all stability devices are effective for increasing muscle activation with highly resistance trained individuals.

Not all studies have reported increased muscle activation with instability devices when the subjects were not highly trained. Behm et al. (2005a) found that there was no significant difference in activation of the trunk musculature during bilateral upper body exercises (chest press, shoulder press) performed on both unstable and stable bases. Anderson and Behm (2004b) also showed no significant increase in activation of the trunk musculature during bench presses performed on the Swiss Ball. Both studies used bilateral contractions of the upper limbs, which may not generate similar disruptive moments seen in unilateral exercises, as both limbs are involved in the movement, allowing the resistance to be maintained directly above the torso and center of gravity. As well, increased load associated with these specific resistance training exercises may serve to distort the instability device and actually provide a more stable platform by flattening at the bottom (Anderson and Behm 2004b).

Results demonstrating similar activation as exhibited by the current subjects may best be explained by Gage et al. (2004). With the body acting as an inverted pendulum (Gage et al. 2004) the center of gravity vacillates constantly. Activation of postural muscles including trunk musculature acts to maintain equilibrium or balance. Since highly resistance trained individuals add further resistance above the center of gravity with many exercises (squats, shoulder press, cleans) there is even a greater stress placed on maintaining the equilibrium of the body's inverted pendulum during these exercises. Hamlyn and Behm (2006), investigated trunk activation under stable and unstable conditions, demonstrating that squats and deadlifts using 45% and 80% of the 1 repetition maximum (1RM) performed on a stable floor elicited a greater activation of the trunk musculature than unstable trunk callisthenic exercises. This study supports Siff (2000) who indicated that the best exercises to stimulate trunk muscles are those, which load the trunk with external resistance such as a squat or deadlift. Thus, some of the instability devices now available such as BOSU balls and Dyna discs may not present sufficient stability challenges to the highly resistance trained individual. The highly resistance trained individual may need to increase their disruptive torque through a combination of load and moderate instability (i.e. squats, deadlifts and cleans).

An established shortcoming of instability training is the lesser ability to load under unstable conditions (Marsden et al. 1983, De Luca and Mambrito 1987, Anderson and Behm 2004b, Behm et al. 2002). Highly resistance trained individuals performing exercises under moderately unstable conditions may not exhibit changes in EMG activity with the exercise. As motor programs are ingrained, the load and its positioning in relation to the centre of gravity may become the formative variables. An investigation into loading using a variety of instability devices may yield further results as to the training effect of these tools. While instability is inherent with free weights, the current importance placed on instability training devices may be over emphasized with individuals who consistently create moderately unstable environments with free weight exercises. Greater degrees of instability such as found with Swiss balls and wobble boards may be necessary in this type of population to increase limb and trunk muscle activation.

Similar to the first two protocols, a moderate degree of instability (BOSU ball) did not produce significant changes in activation in any tested muscles except the soleus during the fatigue protocol or produce a significant change in the rate of fatigue in highly resistance trained individuals. However, stable wall sit conditions elicited greater soleus activation. It is speculated that under unstable conditions with the feet placed upon the moderately unstable BOSU ball, the plantar flexors would not be able to exert similar forces as under stable conditions (Behm et al. 2002). This might force the individuals under unstable conditions to use a greater variety of lower limb muscles (i.e. gastrocnemius, peronei) to lock their lower body into place. The tendency for a greater rate of fatigue with unstable conditions may be related to the additional work of the synergists to cope with the moderate instability.

Conclusion

In conclusion, it has been shown that the use of moderately unstable training devices such as Dyna discs and BOSU balls are not as effective as Swiss balls and wobble boards at increasing activation in the lower body and trunk musculature with highly resistance trained individuals. A determining variable in this research is that all subjects were highly resistance-trained individuals who had extensive experience in the use of heavy free weight resistance and load bearing exercises. The current study tested exercise postures using body weight even though resistance training typically employs the use of greater overload. An investigation into the EMG activity associated with these posture and devices under loaded conditions may provide definitive answers as to the effectiveness of these tools. Moreover, an investigation into the effectiveness of training with instability tools such as the wobble board, Swiss ball, Dyna disc and BOSU ball in a less highly trained population, which may benefit from instability devices would provide useful insight. This may extend to populations who seek to rehabilitate muscle without harboring external load, which may amplify injury or dysfunction.

Figures 3.1 -3.18

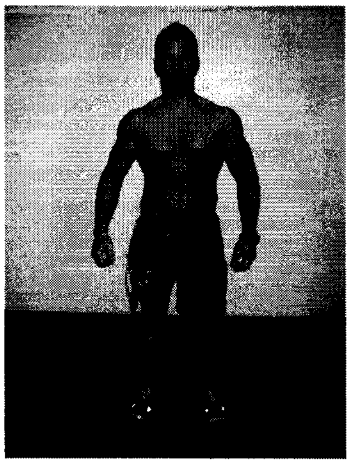


Fig 3.1

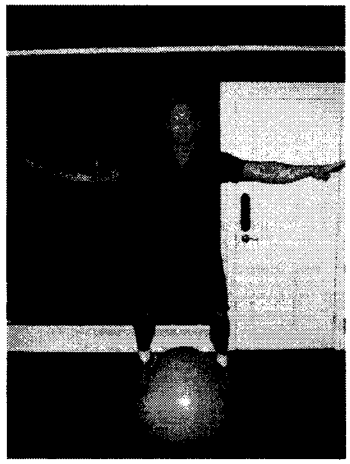


Fig 3.2

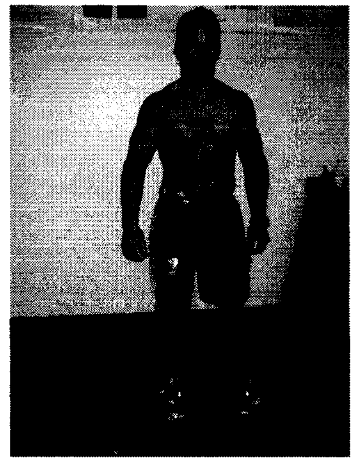
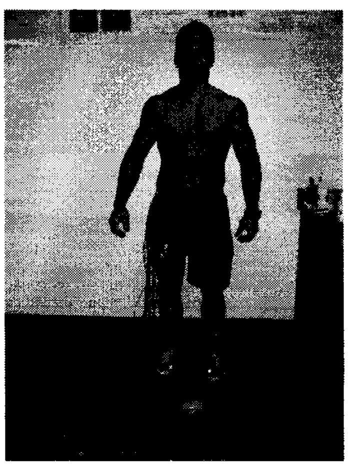


Fig 3.3



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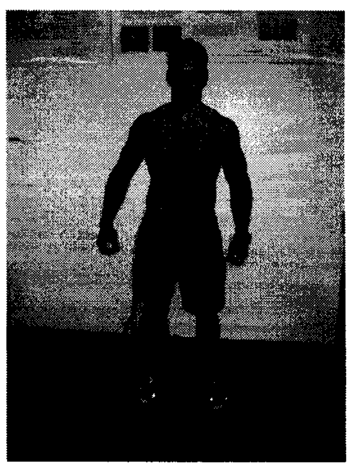

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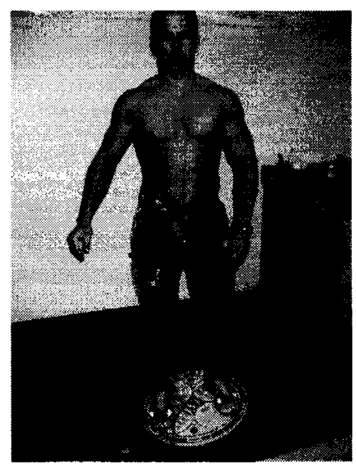

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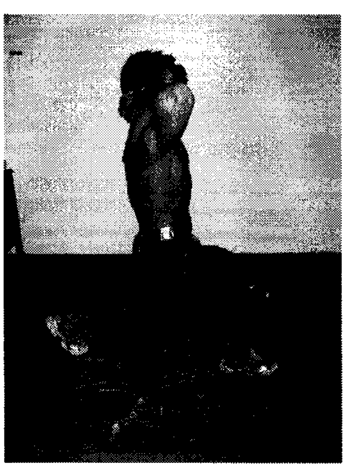

Fig 3.7

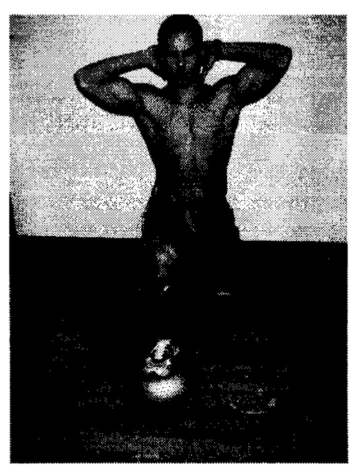

Fig 3.8

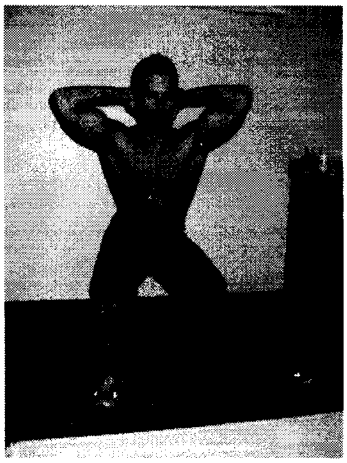

Fig 3.9

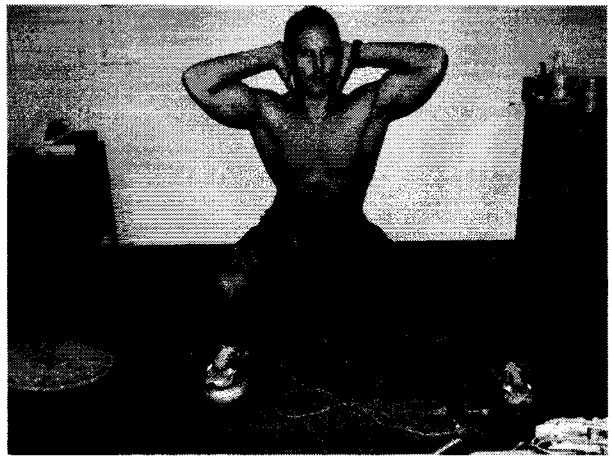

Fig 3.10

Fig 3.11

Fig 3.12

Fig 3.13

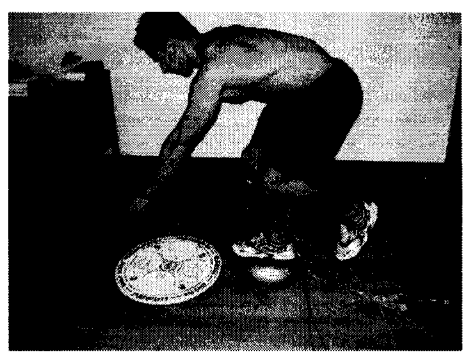

Fig 3.14

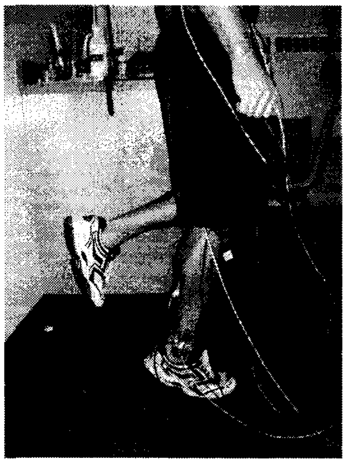

Fig 3.15

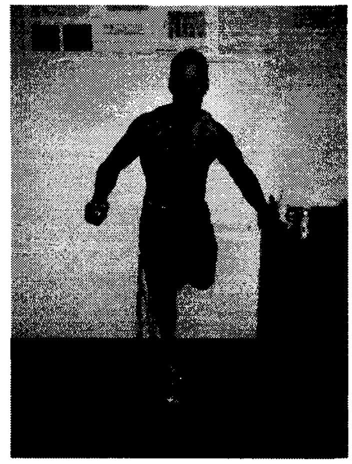

Fig 3.16

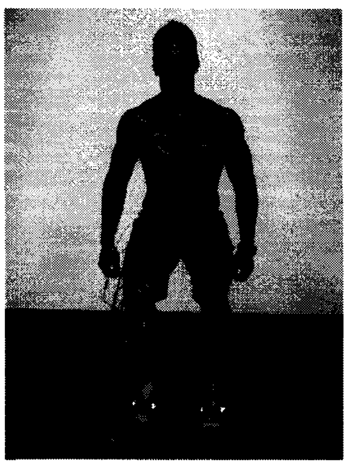

Fig 3.17

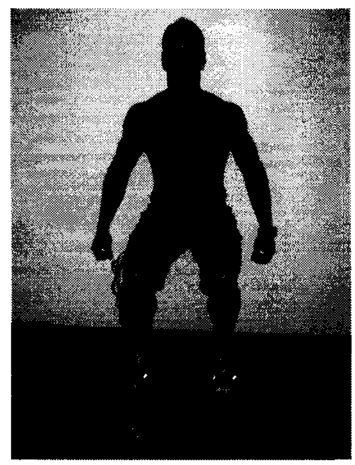

Fig 3.18

Appendix

Table 3.6: 1 way ANOVA main effects for iEMG (mV.s) for the standing posture. Acronyms: LSES: lumbo-sacral erector spinae, LAS: lower abdominals, RF: rectus femoris, BF: biceps femoris

Main Effect	Variables	Muscles	Degrees of Freedom (df)	F ratio	p value
Experiment 1 Stand	Platforms: (BOSU up, BOSU down, Dyna disc, Swiss Ball, Wobble Board, Floor)	LSES	5	15.1	$p<0.0001$
		LAS	5	2.64	$p=0.29$
		RF	5	8.86	$p<0.0001$
		BF	5	12.53	$p<0.0001$
		Soleus	5	3.84	$p=0.0038$

Table 3.7: 1 way ANOVA main effects for iEMG (mV.s) for the squatting posture.

Main Effect	Variables	Muscles	Degrees of Freedom (df)	F ratio	p value
Experiment 1 Squat	Platforms: (BOSU up, BOSU down, Dyna disc, Swiss Ball, Wobble Board, Floor)	LSES	5	1.56	$p=0.18$
		LAS	5	5.67	$p=0.0002$
		RF	5	0.46	$p=0.80$
		BF	5	1.58	$p=0.18$
		Soleus	5	3.84	$p<0.0001$

Table 3.8: Integrated EMG main effect descriptions for the three experiments. Acronyms: LSES. Lumbo-sacral erector spinae, LAS: lower abdominals, RF, rectus femoris, BF: biceps femoris

Main Effect	Variables	Collapsed over:	Muscles	Degrees of Freedom (df)	F ratio	p value
Experiment 2 Stability	Platforms: (Dyna Disc and Floor)	Exercises: (Static Forward Lunge, Static Side Lunge, 1 Leg Hip extension, 1 Leg Reach, Calf Raises)	LSES	1,4	0.15	p=0.69
			LAS	1,4	0.03	p=0.86
			RF	1,4	0.03	p=0.86
			BF	1,4	3.82	p=0.06
			Soleus	1,4	0.04	p=0.85
Experiment 2 Exercises	Exercises: (Static Forward Lunge, Static Side Lunge, 1 Leg Hip extension, 1 Leg Reach, Calf Raises)	Platforms: (Dyna Disc and Floor)	LSES	4,1	8.4	p=0.0001
			LAS	4,1	2.29	p=0.06
			RF	4,1	23.72	p<0.0001
			BF	4,1	65.22	p<0.0001
			Soleus	4,1	85.39	p<0.0001
Experiment 3 Platforms	Platforms: (BOSU up and Floor)	Contractions: 1^{st}, 1/3 2/3 final contraction	LSES	1,3	0.15	p=0.69
			LAS	1,3	0.56	p=0.81
			RF	1,3	0.83	p=0.37
			BF	1,3	2.94	p=0.1
			Soleus	1,3	5.44	p=0.03
Experiment 3 Contractions	Contractions: 1^{st}, 1/3 2/3 final contraction	Platforms: (BOSU up and Floor)	LSES	3,1	0.54	p=0.65
			LAS	3,1	5.04	p=0.003
			RF	3,1	2.6	p=0.06
			BF	3,1	5.91	p=0.001
			Soleus	3,1	6.33	p=0.0008

Table 3.9: Time to Fatigue

Experiment 3 Time to fatigue	Platforms: BOSU up and Floor	1 way ANOVA	1	3.29	p=0.09

Interactive effects. iEMG measures include means ± standard deviations.

Table 3.10: Experiment 1, Lumbo-sacral erector spinae iEMG (mV.s)
$F(1,4) = 15.1$ $p<0.0001$

	Standing	Squatting
BOSU up	0.18 ± 0.01	0.46 ± 0.25
BOSU down	0.18 ± 0.09	0.68 ± 0.82
Dyna Disc	0.21 ± 0.15	0.52 ± 0.34
Swiss Ball	0.33 ± 0.11	0.42 ± 0.22
Wobble Board	0.32 ± 0.16	0.80 ± 0.82
Floor	0.1 ± 0.06	0.76 ± 0.90

Table 3.11: Experiment 1, Lower abdominals iEMG (mV.s)
$F(1,4) = 2.6$ $p=0.03$

	Standing	Squatting
BOSU up	0.12 ± 0.08	0.07 ± 0.06
BOSU down	0.12 ± 0.07	0.09 ± 0.07
Dyna Disc	0.13 ± 0.09	0.08 ± 0.09
Swiss Ball	0.17 ± 0.11	0.18 ± 0.14
Wobble Board	0.18 ± 0.13	0.18 ± 0.15
Floor	0.12 ± 0.07	0.11 ± 0.08

Table 3.12: Experiment 1, Rectus femoris iEMG (mV.s)
$F(1,4) = 8.9$ $p<0.0001$

	Standing	Squatting
BOSU up	0.07 ± 0.05	0.51 ± 0.50
BOSU down	0.07 ± 0.05	0.49 ± 0.44
Dyna Disc	0.08 ± 0.06	0.55 ± 0.50
Swiss Ball	0.19 ± 0.15	0.50 ± 0.32
Wobble Board	0.12 ± 0.10	0.42 ± 0.30
Floor	0.02 ± 0.03	0.48 ± 0.31

Table 3.13: Experiment 1, Biceps Femoris iEMG (mV.s)
$F(1,4) = 12.5$ $p<0.0001$

	Standing	Squatting
BOSU up	0.06 ± 0.04	0.07 ± 0.05
BOSU down	0.07 ± 0.04	0.08 ± 0.05
Dyna Disc	0.05 ± 0.04	0.06 ± 0.07
Swiss Ball	0.10 ± 0.05	0.07 ± 0.03
Wobble Board	0.14 ± 0.09	0.10 ± 0.08
Floor	0.04 ± 0.03	0.05 ± 0.02

Table 3.14: Experiment 1, Soleus iEMG (mV.s)
$F(1,4) = 3.8$ $p=0.004$

	Standing	Squatting
BOSU up	0.33 ± 0.45	0.17 ± 0.11
BOSU down	0.36 ± 0.45	0.21 ± 0.13
Dyna Disc	0.32 ± 0.45	0.23 ± 0.14
Swiss Ball	0..43 ± 0.35	0.28 ± 0.19
Wobble Board	0.58 ± 0.50	0.41 ± 0.30
Floor	0.28 ± 0.45	0.13 ± 0.08

Table 3.15: Experiment 2, Lumbo-sacral erector spinae iEMG (mV.s)
F(1,4) = 1.6; p=0.17

	Dyna Disc	Floor
Static Forward Lunge	0.38 ± 0.20	0.37 ± 0.22
Static Side Lunge	0.42 ± 0.26	0.52 ± 0.24
1 Leg Hip extension	0.57 ± 0.26	0.66 ± 0.32
1 Leg Reach	0.26 ± 0.26	0.17 ± 0.11
Calf Raises	0.66 ± 0.81	0.43 ± 0.22

Table 3.16: Experiment 2, Lower abdominal iEMG (mV.s)
F(1,4) = 0.24; p=0.92

	Dyna Disc	Floor
Static Forward Lunge	0.16 ± 0.25	0.11 ± 0.13
Static Side Lunge	0.15 ± 0.18	0.13 ± 0.23
1 Leg Hip extension	0.20 ± 0.13	0.22 ± 0.14
1 Leg Reach	0.27 ± 0.25	0.28 ± 0.27
Calf Raises	0.27 ± 0.29	0.36 ± 0.87

Table 3.17: Experiment 2, Rectus femoris iEMG (mV.s)
F(1,4) = 1.06; p=0.38

	Dyna Disc	Floor
Static Forward Lunge	0.34 ± 0.26	0.47 ± 0.51
Static Side Lunge	0.72 ± 0.70	0.72 ± 0.64
1 Leg Hip extension	0.03 ± 0.01	0.04 ± 0.03
1 Leg Reach	0.19 ± 0.30	0.17 ± 0.14
Calf Raises	0.68 ± 0.47	0.48 ± 0.38

Table 3.18: Experiment 2, Biceps femoris iEMG (mV.s)
F(1,4) = 1.84; p=0.13

	Dyna Disc	Floor
Static Forward Lunge	0.17 ± 0.15	0.15 ± 0.21
Static Side Lunge	0.12 ± 0.07	0.11 ± 0.06
1 Leg Hip extension	0.93 ± 0.45	0.69 ± 0.37
1 Leg Reach	0.15 ± 0.07	0.14 ± 0.10
Calf Raises	0.18 ± 0.11	0.12 ± 0.05

Table 3.19 Experiment 2, Soleus iEMG (mV.s)
F(1,4) = 0.39; p=0.82

	Dyna Disc	Floor
Static Forward Lunge	0.35 ± 0.39	0.31 ± 0.24
Static Side Lunge	0.35 ± 0.33	0.37 ± 0.25
1 Leg Hip extension	0.25 ± 0.13	0.17 ± 0.15
1 Leg Reach	1.15 ± 0.39	1.20 ± 0.48
Calf Raises	0.64 ± 0.42	0.61 ± 0.39

Table 3.20: Experiment 3, Fatigue-related iEMG (mV.s) of Lumbo-sacral Erector Spinae
F(1,3) = 0.27; p=0.84

	BOSU up	Floor
First contraction	0.60 ± 0.33	0.53 ± 0.37
Contraction at 1/3 duration	0.53 ± 0.25	0.48 ± 0.21
Contraction at 2/3 duration	0.56 ± 0.26	0.50 ± 0.21
Final contraction	0.57 ± 0.30	0.60 ± 0.33

Table 3.21: Experiment 3, Fatigue-related iEMG (mV.s) of Lower Abdominals
$F(1,3) = 0.23$; $p=0.87$

	BOSU up	Floor
First contraction	0.13 ± 0.19	0.16 ± 0.22
Contraction at 1/3 duration	0.10 ± 0.13	0.11 ± 0.09
Contraction at 2/3 duration	0.12 ± 0.14	0.13 ± 0.11
Final contraction	0.19 ± 0.19	0.19 ± 0.16

Table 3.22: Experiment 3, Fatigue-related iEMG (mV.s) of Rectus femoris
$F(1,3) = 1.58$; $p=0.2$

	BOSU up	Floor
First contraction	0.41 ± 0.23	0.40 ± 0.19
Contraction at 1/3 duration	0.54 ± 0.32	0.39 ± 0.22
Contraction at 2/3 duration	0.52 ± 0.35	0.39 ± 0.20
Final contraction	0.55 ± 0.37	0.45 ± 0.19

Table 3.23: Experiment 3, Fatigue-related iEMG (mV.s) of Biceps femoris
$F(1,3) = 2.18$; $p=0.001$

	BOSU up	Floor
First contraction	0.07 ± 0.05	0.06 ± 0.02
Contraction at 1/3 duration	0.11 ± 0.08	0.06 ± 0.03
Contraction at 2/3 duration	0.11 ± 0.09	0.06 ± 0.03
Final contraction	0.14 ± 0.09	0.07 ± 0.03

Table 3.24: Experiment 3, Fatigue-related iEMG (mV.s) of Soleus
$F(1,3) = 6.04$; $p=0.001$

	BOSU up	Floor
First contraction	0.21 ± 0.18	0.33 ± 0.30
Contraction at 1/3 duration	0.25 ± 0.24	0.40 ± 0.28
Contraction at 2/3 duration	0.21 ± 0.18	0.49 ± 0.28
Final contraction	0.22 ± 0.18	0.62 ± 0.34

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