Journal of Strength and Conditioning Research, 2007, 21(2), 000–000

ELECTROMYOGRAPHIC ACTIVITY OF THE TRUNK STABILIZERS DURING STABLE AND UNSTABLE BENCH PRESS

JEFF NORWOOD, GREGORY S. ANDERSON, MICHAEL GAETZ, AND PETER TWIST

Affiliation Addresses

ABSTRACT. Norwood, J., G.S. Anderson, M. Gaetz, and P. Twist. Electromyographic activity of the trunk stabilizers during stable and unstable bench press. *J. Strength Cond. Res. 21(2):000–000. 2007.*—The purpose of this study was to investigate the effectiveness of instability training in the recruitment of core stabilizing muscles during dynamic multijoint movement. Surface electromyography (EMG) was measured from 6 muscles (latissimus dorsi, rectus abdominus, internal obliques, erector spinae, and soleus) while subjects performed a 9.1-kg bench press on stable and unstable surfaces. There were 4 exercises in total: (a) stable surfaces for shoulders and feet, (b) upper-body instability, (c) lower-body instability, and (d) dual instability. Five seconds of EMG were recorded during each bench press and were subsequently smoothed with root mean squares calculated for the entire time-series. A repeated-measures analysis of variance (ANOVA) was used to test overall differences between exercise conditions for each muscle. Paired equal variance *t*-tests with a stepwise Bonferroni correction for multiple contrasts ($\alpha = 0.05/$total number of contrasts) were performed for muscles with significant repeated-measures ANOVA results. The results show significant increases in EMG with increasing instability. Specifically, the dual instability bench press resulted in the greatest mean muscle activation of the 3 stability conditions, with single instability conditions being significantly greater than the stable condition. This pattern of results is consistent with the position that performing the bench press in a progressively unstable environment may be an effective method to increase activation of the core stabilizing musculature, while the upper- and lower-body stabilizers can be activated differentially depending on the mode of instability.

KEY WORDS. single instability, dual instability, weight training

INTRODUCTION

In recent years, many stability challenging devices have been introduced to condition the core musculature. Devices such as the stability and BOSU balls have been widely prescribed for use with athletes to serve this purpose. However, despite the prevalence of its prescription, little research has been done to assess the effectiveness of instability training in the recruitment of core stabilizing muscles during dynamic multijoint movement.

Marjorie (2000) defined the core as the physical area spanning the entire trunk area extending from the rib cage to the pelvis. The upper core comprises the rib cage, where its positioning has a direct influence on the function of the shoulder girdle and thus on the mobility and control of the upper appendages. The bottom third of the core comprises the pelvis, where its location affects the function and stability of the lower appendages. The middle third of the core consists primarily of the tissues in the abdominal and low-back regions. It has been suggested that imbalances in the static postural muscles of the middle third of the core can alter the position, stability, and neuromuscular control of the levers through the entire link system and, therefore, the mechanical control of the region. Through the middle third of the core the positioning of the pelvis is linked to the rib cage and thus the shoulder girdle. This indicates that the mechanical and neuromuscular control of the core musculature can greatly affect the mobility and function of the extremities (Marjorie, 2000) and, thus, may be an appropriate exercise prescription variable.

Two studies by Behm and colleagues (5, 8) found increased electromyographic (EMG) activity in the trunk stabilizers when subjects were performing trunk strengthening and chest and shoulder press exercises on unstable platforms when incorporating unilateral limb movement. Likewise, Anderson and Behm (2) found significant increases in trunk stabilizer and postural muscle activity when introducing progressive instability during squat performance. That same study, however, found only negligible increases in prime mover muscle activation with the introduction of unstable platforms. The authors suggested that the use of instability training may be more beneficial for the activation of stabilizing muscles, rather than the prime movers. This being the case, instability training should be beneficial as a rehabilitation tool where less resistance would pass through a muscle or joint system for an equivalent level of muscle activation, while stressing trunk and joint stabilizers during exercise.

The use of unstable training surfaces is hypothesized to increase the activation of stabilizing musculature. However, to date no research has compared muscle recruitment during differential limb segment instability. Research to date has typically encouraged destabilization of the shoulders, by placing them on an unstable surface such as a stability ball, or of the trunk, by having subjects sit on a stability ball. The purpose of this study was to examine differences in EMG muscle activation of the trunk stabilizers while subjects were performing bench press movements in stable, single, and dual instability environments. Instability was created by placing either the shoulders or the feet or both on an unstable surface. It was hypothesized that EMG muscle activation values in the trunk and upper-/lower-body stabilizers would increase as the subject progressed from a stable environment to one involving dual instability.

METHODS

Experimental Approach to the Problem

Subjects performed a bench press under 4 conditions. Muscle activity was recorded during each bench press using standard EMG procedures. Root mean squares (RMS) were calculated for each muscle and for each condition to quantify muscle activation. Using this procedure differences in muscle activity could be determined during each of the stable, single, and dual instability conditions.

Subjects

Subjects included 10 male and 5 female elite conditioning
?3 coaches and/or personal trainers with an average of 8.4 years of experience. Subjects had an average of 10.4 years of training experience with the bench press, 5.8 years of training experience with the stability ball, and 1.3 years of training experience with the BOSU Balance Trainer. All subjects were screened for health issues that would limit participation using a Physical Activity Readiness Questionnaire. Each participant signed an informed consent prior to participating in the study, which was approved by the institution's research ethics board.

Instrumentation

Surface EMG recording locations were measured for the
?4 following 6 muscle groups ([Basmajian, Blumstein, 1980], 2, 6, 11); latissimus dorsi (10 cm lateral to T9), rectus abdominus (muscle bellies lateral to the umbilicus), internal obliques (2.5 cm medial from the anterior superior iliac spine), erector spinae (2 cm lateral to L5-S1, bicep femoris [midbelly of the long head]), and soleus (midpoint between medial malleolus and medial condyle of the tibia). Each EMG site was shaved, cleansed with alcohol, and abraded with interelectrode distances of approximately 2.5 cm positioned parallel to the orientation of the fibers being measured.

Surface EMG activity was recorded using Grass Model
?5 10A amplifiers with a digital interface. Electrical activity was recorded using 17 silver–silver chloride disposable electrodes applied to the skin's surface (16 EMG electrodes and a ground). Electrode impedances were maintained below 5 kOhms. Filter settings were 10 Hz for the low filter and 1,000 Hz for the high filter. The recording epoch was 5 seconds at a sample rate of 2,000 Hz. All data were digitized using a National Instruments AT-MIO-16F-5 A-D card and stored on a CD. The data were smoothed using a 10-point moving average, and RMS were calculated for time-series using a Microsoft EXCEL spreadsheet.

Procedures

Each subject was instructed to perform a bench press movement using a 9.1-kg steel lifting bar in 4 different "stability" conditions (see Figure 1): back on the bench, feet on the floor (stable); back on the bench, feet on the BOSU platform (single instability—lower body [LBI]); back on stability ball, feet on the floor (single instability—upper body [UBI]); and back on stability ball, feet on BOSU platform (dual instability [DI]).

In the stable condition each subject was positioned supine on the bench with knees flexed to 90° and feet parallel on the floor. During single instability of the lower body the subjects were instructed to hold a bridge position with their shoulders supported by the bench with enough clearance for the latissimus dorsi electrodes to be free of the bench's proximal edge (the ninth thoracic vertebra was off of the bench). The subject's feet were positioned on the platform side of the BOSU ball with knees flexed at 90°. During all bridging postures subjects were reminded to elevate their pelvic girdle to be approximately parallel with their shoulders.

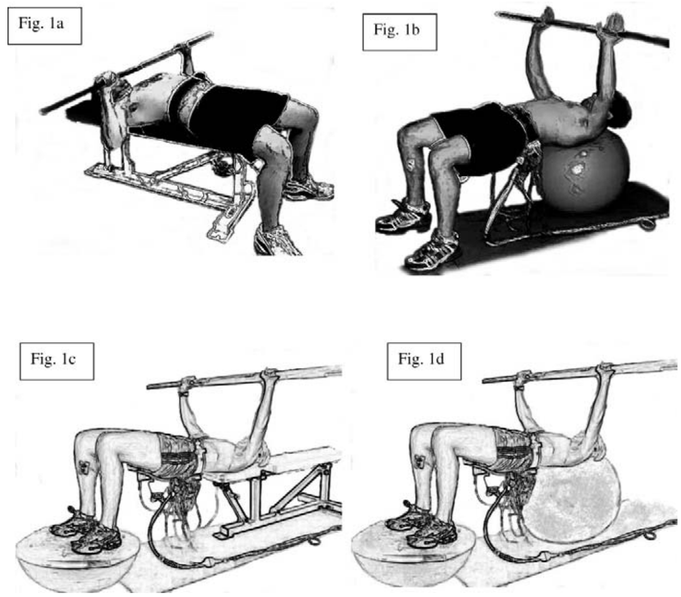

FIGURE 1. a, Bench press; b, upper-body instability; c, lower-body instability; and d, dual instability.

Single instability of the upper body was achieved by asking each subject to perform a bridge using a stability ball that supported the shoulders across the first 8 thoracic vertebrae. For this exercise, the subject's knees were flexed at 90° with feet placed on the floor.

Instability of both the upper and lower body (dual instability) was tested with each subject performing a bridge with the shoulders on the stability ball (in a similar manner as described for the single upper-body instability) and feet placed on the platform side of the BOSU ball (in a similar manner as described for the single lower-body instability).

Each subject performed one repetition for each test condition. A metronome was used to count down trial initiation and for timing the eccentric and concentric portions of the exercise. Subjects started the bench press with elbows fully extended, maintaining a grip distance slightly wider than their shoulders' width. The bar was then lowered to their chest and then pressed back to full elbow extension on a 2-1-2 count. The timing sequence involved 2 seconds for the eccentric phase, a 1-second hold, and 2 seconds for the "up" concentric phase while returning the bar to the start position. Electromyographic recording was initiated at the end of a 3-second countdown and continued over the full 5 seconds required to complete the exercise. Subjects were asked to repeat trials when there was a miscommunication or an electrical artefact or when the subject could not maintain his balance long enough to complete the entire exercise. The minimum time interval between trials was 30 seconds.

Statistical Analyses

Statistical analyses for all RMS EMG measures used a repeated-measures analysis of variance (ANOVA; General Linear Model ANOVA, version 12.9; SPSS, Inc., Chicago, IL) to test for overall differences between exercise conditions for each muscle. The Huynh-Feldt adjustment for violations of sphericity was utilized, and degrees of freedom reported were Huynh-Feldt adjusted based on the value of epsilon. Linear effects for each muscle across exercise conditions were analyzed to determine the presence of significant trends in the data.

TABLE 1. Subject characteristics.

Gender*	Age (y)	Height (cm)	Weight (kg)	Coaching experience (y)	Training experience (y)
F	24	167.6	56.8	3	18.0
F	38	172.7	58.2	4	15.0
F	23	174.0	64.1	3	10.0
F	29	165.0	67.0	8	16.0
F	38	172.7	62.7	15	12.0
M	31	169.0	68.0	3	10.0
M	33	177.8	95.5	5	5.0
M	27	175.3	75.0	5	14.0
M	40	182.9	90.9	20	20.0
M	21	180.3	83.2	3	2.0
M	37	177.8	88.6	12	20.0
M	27	185.4	85.9	7	10.0
M	25	184.2	102.3	6	10.0
M	23	193.0	109.0	6	6.0
M	24	193.0	88.6	2	6.0
Mean	29.3	178.1	79.7	7.1	10.4
SD	6.4	8.5	16.5	5.1	5.9

* F = female; M = male.

In addition, paired equal-variance t-tests with a stepwise Bonferroni correction for multiple contrasts (α = 0.05/total number of contrasts) were performed for muscles with significant repeated-measures ANOVA results. This methodology included corrections to α for all post-hoc contrasts that were performed, resulting in a conservative estimate of significant mean differences between groups. For the statistically significant t-tests, effect sizes were computed using the method proposed by Cohen (10): d = the mean difference between the control group and the depressed group divided by the pooled *SD*. Technically, the pooled *SD* is a weighted *SD* computed by multiplying the individual group *SD*s by the proportion of subjects each group contributes to the total sample and then summing these 2 products. Effect sizes of 0.2 represent "small effects," effect sizes of 0.5 "medium effects," and effect sizes of 0.8 "large effects."

Finally, intraclass correlations were performed for the 4 experimental conditions within muscle using a 2-way mixed model. Rejection of the null hypothesis indicated reliability, and p values were considered significant when $\leq$0.05.

RESULTS

?6 The mean and *SD*s for RMS EMG for each muscle and exercise condition are presented in Table 2. A total of 23 (of the 36 possible within muscle) post-hoc contrasts were performed. Contrasts not performed included those with no significant ANOVA results and those in which the unstable surface was distant from the muscle investigated (e.g., the single instability–lower limb condition was not included for latissimus dorsi contrasts). This resulted in an initial correction factor of 0.05/23 (p = 0.0022) for the first pass of the stepwise Bonferroni. Ten of the initial 23 post-hoc contrasts were significant at this level. The correction factor for the second pass was adjusted to 0.05/14 (p = 0.0038). One additional paired contrast was deemed significant. No other paired contrasts reached significance following the third pass (0.05/12).

Latissimus Dorsi

The repeated-measures ANOVA revealed a significant difference for RMS EMG activity from latissimus dorsi ($F_{1,1.35}$ = 5.303, p = 0.024). Significant linear effects were also observed ($F_{1,3}$ = 5.360, p = 0.036), indicating that with increasing instability, RMS values increased (see Figure 1). Three paired post-hoc contrasts were performed for this muscle: the stable condition vs. single instability upper, the stable condition vs. dual instability, and single instability upper vs. dual instability. No significant paired post-hoc contrasts were observed following the stepwise Bonferroni correction. The intraclass correlation for latissimus dorsi was r = 0.266 ($F_{14,42}$ = 1.363, p = 0.214).

Rectus Abdominus

The repeated-measures ANOVA did not reveal a significant difference for RMS EMG activity from rectus abdominus ($F_{1,1.46}$ = 1.870, p = 0.118). No significant linear effects were observed ($F_{1,3}$ = 0.125, p = 0.650). Although there were no significant effects for the ANOVA or trend analysis, it was observed that the mean values for single upper and lower instability were much lower than RMS for dual instability. Significant post-hoc paired contrasts were observed following the stepwise Bonferroni correction for single instability–lower vs. dual instability [$t(14)$ = −3.58; p = 0.003; d = 1.69, very large effect] and single instability–upper vs. dual instability [$t(14)$ = −4.24; p = 0.001; d = 1.57, very large effect]. The intraclass correlation for rectus abdominus was r = 0.595 ($F_{14,42}$ = 2.469, p = 0.012).

Internal Oblique

The repeated-measures ANOVA revealed a significant difference for RMS EMG activity from the internal oblique ($F_{1,1.59}$ = 12.43, p = 0.002). Significant linear effects were also observed ($F_{1,3}$ = 15.95, p = 0.001), indicating that with increasing instability, RMS values increased (see Figure 1). Significant post-hoc paired contrasts were observed following the stepwise Bonferroni correction for the stable condition vs. single instability–lower [$t(14)$ = −6.22; p = 0.000; d = 2.28, very large effect], the stable condition vs. single instability–upper [$t(14)$ = −6.50; p = 0.000; d = 2.42, very large effect], and the stable condition vs. dual instability [$t(14)$ = −4.03; p = 0.001; d = 1.9, very large effect]. No significant paired

TABLE 2. Mean (±*SD*) of root mean squares (RMS) electromyography (EMG) for 15 subjects by condition and muscle.

Muscle	Stable	Single instability (lower body)	Single instability (upper body)	Dual instability
Latissimus dorsi	9.32 (8.97)	8.53 (5.05)	11.76 (6.71)	26.14 (25.24)
Rectus abdominus	7.90 (8.15)	6.47 (2.92)	5.70 (3.70)	9.19 (4.06)
Internal obliques	7.59 (5.10)	33.54 (17.64)	33.21 (16.05)	74.07 (64.85)
Erector spinae	9.70 (9.83)	11.64 (12. 63)	7.23 (4.33)	23.24 (27.63)
Soleus	3.15 (1.24)	9.54 (4.76)	9.06 (11.09)	19.78 (11.27)
Biceps femoris	4.01 (2.09)	24.49 (16.06)	10.25 (4.90)	40.57 (24.66)

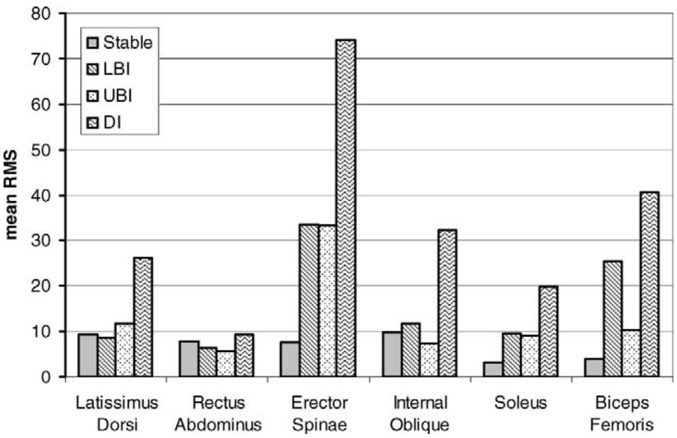

FIGURE 2. Mean muscle activation for each stability condition UBI = upper-body instability; LBI = lower-body instability; DI = dual instability.

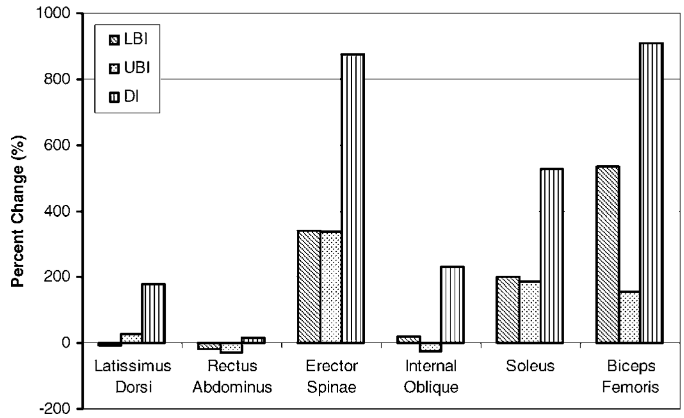

FIGURE 3. Electromyographic (EMG) root mean squares (RMS) expressed as a percent change from stable condition values. UBI = upper-body instability; LBI = lower-body instability; DI = dual instability.

post-hoc contrasts were observed following the stepwise Bonferroni correction for single instability–lower vs. single instability–upper, single instability–lower vs. dual instability, and single instability–upper vs. dual instability. The intraclass correlation for the internal obliques was $r = 0.545$ ($F_{14,42} = 2.197$, $p = 0.025$).

Erector Spinae

The repeated-measures ANOVA revealed a significant difference for RMS EMG activity from the erector spinae ($F_{1,1.65} = 9.19$, $p = 0.002$). Significant linear effects were also observed ($F_{1,3} = 8.47$, $p = 0.011$), indicating that with increasing instability, RMS values increased (see Figure 1). Of the 6 possible within-muscle contrasts performed, the only significant post-hoc paired contrast following the stepwise Bonferroni correction was for single instability–upper vs. dual instability [$t(14) = -3.75$; $p = 0.002$; $d = 1.57$, very large effect]. Intraclass correlation for erector spinae was $r = 0.661$ ($F_{14,42} = 2.953$, $p = 0.003$).

Biceps Femoris

The repeated-measures ANOVA revealed a significant difference for RMS EMG activity from the biceps femoris ($F_{1,2.65} = 13.84$, $p = 0.000$). Significant linear effects were also observed ($F_{1,3} = 28.08$, $p = 0.000$), indicating that with increasing instability, RMS values increased (see Figure 1). Significant post-hoc paired contrasts were observed following the stepwise Bonferroni correction for the stable condition vs. single instability–lower [$t(14) = -5.14$; $p = 0.000$; $d = 2.13$, very large effect], the stable condition vs. dual instability [$t(14) = -5.73$; $p = 0.000$; $d = 2.66$, very large effect], and the single instability–lower vs. dual instability [$t(14) = -3.91$; $p = 0.001$; $d = 1.28$, very large effect]. The intraclass correlation for biceps femoris was $r = 0.558$ ($F_{14,42} = 2.262$, $p = 0.021$).

Soleus

The repeated-measures ANOVA revealed a significant difference for RMS EMG activity from the soleus ($F_{1,1.81} = 19.67$, $p = 0.000$). Significant linear effects were also observed ($F_{1,3} = 25.34$, $p = 0.000$), indicating that with increasing instability, RMS values increased (see Figure 1). Significant post-hoc paired contrasts were observed following the stepwise Bonferroni correction for the stable condition vs. single instability–lower [$t(14) = -4.96$; $p = 0.000$; $d = 2.37$, very large effect] and the stable condition vs. dual instability [$t(14) = -5.70$; $p = 0.000$; $d = 2.73$, very large effect]. No significant paired post-hoc contrasts were observed following the stepwise Bonferroni correction for single instability–lower vs. dual instability. The intraclass correlation for soleus was $r = 0.252$ ($F_{14,42} = 1.337$, $p = 0.228$).

Percent Change

The increase in muscle recruitment can be seen by comparing EMG results from the stable condition to those of the single and dual instability conditions. Changes, expressed as a percent change from the stable condition ([RMS $\times/\div$ RMS stable] $-$ 1) are presented in Figure 2. ?7
These values show increased muscle activity and not absolute EMG RMS values and, therefore, represent the increased recruitment of each muscle.

DISCUSSION

Researchers and trainers who promote training on unstable platforms claim that utilizing equipment like the stability ball and the BOSU Balance Trainer provides a greater stress to the overall musculature (8, [Sheth, 1997], 12, 13, 21). It has been hypothesized that perform- ?8
ing exercise in an unstable environment stresses the synergistic and stabilizing muscles around a joint system for any given movement, providing a more specific and functional form of training. The present results support this position.

The greatest mean muscle activation of the 3 stability conditions was found when subjects were performing the dual instability bench press. Single instability, with upper- and lower-body instability grouped together, yielded significantly greater mean muscle activation than the stable condition. These results indicate that performing the bench press in a progressively unstable environment may prove an effective means to increase activation of the core stabilizing musculature and, depending on the mode of instability, the upper- and lower-body stabilizers as well. These results differ from those of Anderson and Behm (1), who found no significant difference in overall EMG activity between stable and unstable bench press protocols. However, this finding could be due to the fact that Anderson and Behm only measured EMG activity in the bench press prime movers and the rectus abdominus. The prime movers have been reported to play an insignificant role in stabilization and are thus not affected by the unstable environments. The use of unstable platforms in resistance training may be more beneficial for the trunk stabilizers than for the prime movers (2, 6, 7). The same phenomena was observed across the shoulder by McCaw and Friday (17), who found that greater muscular activity was present during a free-weight shoulder press compared to during a machine press, with differences most predominant in the shoulder stabilizers.

When comparing the difference between upper- and lower-body instability and stable conditions, only single instability of the lower body produced significantly different activation values. The majority of muscles, including the rectus abdominus and the internal obliques, showed higher EMG activation during lower-body instability. The only muscle group that showed a higher activation during upper-body instability was the latissimus dorsi, which makes sense, since this muscle's contribution to the movement takes place in the upper body, where the body interfaces with the mode of instability. These results are similar to those of Behm et al. (7), who found no significant difference in the muscle activation of the prime movers and rectus abdominus during performance of a bench press in stable and upper-body instability conditions.

Behm et al. (7) hypothesized that by positioning the athlete's center of mass outside the center of the ball during the bench press performed with the shoulders on a stability ball a stronger destabilizing torque would be created and would require a greater activation of the trunk stabilizers to maintain stability. Instability is not only achieved through the use of unstable bases (stability ball, BOSU) and resistance modality (free-weights) but also with destabilizing torques. Behm et al. (7) found that unbalanced, unilateral movement of resistance by a single arm outside the base of support would lead to a destabilizing torque that must be countered by contraction of the contralateral trunk musculature during chest and shoulder presses. Dual instability and lower-body instability in the present study may have increased the destabilizing torque through the core and, thus, increased the activation of the core musculature.

The most challenging exercise performed in the present study was the dual instability bench press, which recorded the greatest muscle activation for each of the exercise conditions and the greatest percent change in muscle activity. This is the first study to record EMG values while using a dual instability protocol. The majority of individual muscles showed a statistically significant difference in activation in this condition compared to each of the other exercise conditions. The only muscle that did not show a significant change in activation was the rectus abdominus, a result that is in agreement with earlier results (2, 7). This result could be due to the fact that the rectus abdominus acts as a prime mover for trunk flexion, and in each stability and exercise condition for the bench press the trunk maintained a statically extended posture.

Because our population consisted of elite athletes who have prior experience with instability training, the difference in muscle activation across the 3 levels of instability may be diminished because of their proficiency with the equipment. Perhaps in a less well-trained population an even more explicit change in muscle activation with instability modalities would occur. For example, in a less well-trained population there may be a significant increase in muscle activation while using upper-body instability.

PRACTICAL APPLICATIONS

The bench press has been identified as a widely used and effective means of training the upper body. It is a dynamic, multijoint effort that involves adduction of the arm at the shoulder and extension of the forearm at the elbow. The prescription of unstable surfaces has the goal of increasing the muscle activity in lieu of increasing mechanical load. This may provide value for young athletes in training, aging recreationalists, and in-season elite athletes who play several games per week, who cannot always train with low-RM conditions. Furthermore, unstable training may increase the activation of core stabilizers (even during low-RM training), better connecting the kinetic chain so that bench press strength can be expressed on the playing field in closed kinetic chain positions.

In the present study the prescription of instability training during the bench press, whether single or dual, has been shown to increase total muscle activation of the stabilizing musculature, compared to the stable condition. For individuals who do not have the skill, neuromuscular control, or core strength to perform a dual instability bench press, adding single instability of the lower body appears to have a stronger effect on muscle activation than does upper-body instability, especially in the recruitment of trunk musculature and lower leg stabilizers.

REFERENCES

?9

1. ANDERSON, K., AND D.G. BEHM. Maintenance of EMG activity and loss of force output with instability. *Can. J. Appl. Phys.* 28(Suppl.):S27. 2003.
2. ANDERSON, K., AND D.G. BEHM. Trunk muscle activity increases with unstable squat movements. *Can. J. Appl. Phys.* 30:33–45. 2005.
3. BAECHLE, T., AND R. EARLE. *Essentials of Strength Training and Conditioning.* Champaign, IL: Human Kinetics, 2000.
4. BARNETT, C., V. KIPPERS, AND P. TURNER. Effects of variations of the bench press exercise on the EMG activity of five shoulder muscles. *J. Strength Cond. Res.* 9:222–227. 1995.
5. BEHM, D.G. Neuromuscular implications and applications of resistance training. *J. Strength Cond. Res.* 9:264–274. 1995.
6. BEHM, D.G., K. ANDERSON, AND R.S. CURNEW. Muscle force and activation under stable and unstable conditions. *J. Strength Cond. Res.* 16:416–422. 2002.
7. BEHM, D.G., A.M. LEONARD, W.B. YOUNG, W.A.C. BONSEY, AND S.N. MACKINNON. Trunk muscle EMG activity with unstable and unilateral exercises. *Can. J. Appl. Physiol.* 28(Suppl.):S30. 2003.
8. BEHM, D.G., A.M. LEONARD, W.B. YOUNG, W.A.C. BONSEY, AND S.N. MACKINNON. Trunk muscle EMG activity with unstable and unilateral exercises. *J. Strength Cond. Res.* 19:193–201. 2005.
9. BROOKS, D., AND C.C. BROOKS. *Integrated Balance Training: A Programming Guide for Fitness and Health Professionals.* City: DW Fitness, 2002. pp. 4–37. ?10
10. COHEN, J. *Statistical Power Analysis for the Behavioural Sciences* (2nd ed.). Hillsdale, NJ: Erlbaum, 1988.
11. CRAM, J.R., G.S. KASAM, AND J. HOLTZ. *Introduction to Surface Electromyography.* Gaithersburg, MD: Aspen Publishers, 1998.
12. GANTCHEV, G.N., AND D.M. DIMITROVA. Anticipatory postural adjustments associated with arm movements during balancing on unstable support surface. *Int. J. Psychophysiol.* 22:117–122. 1996.
13. IVANENKO, Y.P., Y.S. LEVIK, V.L. TALIS, AND V.S. GURFINKEL. Human equilibrium on unstable support: The importance of feet-support interaction. *Neurosci. Lett.* 235:109–112. 1997.
14. KING, M.A. Core stability: Creating a foundation for functional rehabilitation. *Athl. Ther. Today* 5(2):6–13. 2000.
15. KORNEKI, S., AND V. ZSCHORLICH. The nature of the stabilizing functions of skeletal muscles. *J. Biomech.* 27:215–225. 1994.
16. LEAR, L., AND M. GROSS. An electromyographical analysis of the scapular stabilizing synergists during a push-up progression. *J. Orthop. Sports Phys. Ther.* 146–157. 1998. ?11
17. MCCAW, S.T., AND J.J. FRIDAY. A comparison of muscle activity between a free weight and machine bench press. *J. Strength Cond. Res.* 8:259–264. 1994.
18. RUTHFORD, O.M., AND D.A. JONES. The role of learning and co-ordination in strength training. *Eur. J. Appl. Physiol.* 55:100–115. 1986.

19. SALE, D. Neural adaptation to resistance training. *Med. Sci. Sport Exerc.* 20:135–145. 1988.
20. VERA-GARCIA, F., S. GRENIER, AND S. MCGILL. Abdominal muscle response during curl-ups on both stable and labile surfaces. *Phys. Ther.* 80: 564–569. 2000.
21. WESTER, J., S. JESPERSEN, K. NIELSON, AND L. NEUMANN. Wobble board training after partial sprains of the lateral ligaments of the ankle: A prospective randomized study. *J. Orthop. Sports Phys. Ther.* 332–336. 1996.

Acknowledgments

The authors would like to thank Dr. David Behm (Memorial University of Newfoundland) for his insight into the research question and Brynne Elliott (Research Assistant) for her help in lab preparation, data transformation, and organization.

Address correspondence to Dr. Gregory S. Anderson, Gregory.Anderson@ucfv.ca.