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Effects of Orthoses on Standing Postural Control and Muscle Activity in Children With Cerebral Palsy

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Abstract

Purpose: This exploratory study assessed postural control and muscle activity in children with cerebral palsy while standing barefoot (BF), in prescribed ankle-foot orthoses (AFOs) and in distal control orthoses (DCOs), which stabilized foot-ankle and deliberately aligned the shank.

Methods: This within-subject study evaluated 10 participants, Gross Motor Functional Classification System level III, across the 3 ankle-foot conditions in: (1) static standing duration and (2) modified Clinical Test of Sensory Interaction on Balance with electromyography (EMG) on 7 muscles.

Results: Participants had significantly decreased center of gravity (COG) velocity sway in DCO versus BF and AFO, decreased loss of balance (LOB), and increased standing for DCO versus BF. DCO had minimal effect on EMG activity.

Conclusions: DCO provided significant stabilizing effects on COG sway velocity, standing duration, and LOB. DCO may be effective in balance training. It is unclear whether benefit was derived from stabilization of the ankle joint, the resultant shank alignment, or both.

Keywords

ankle-foot orthoses; cerebral palsy; postural control; segmental kinematics

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INTRODUCTION

For children with cerebral palsy (CP), Gross Motor Functional Classification System (GMFCS) level III (ie, need a hand-held mobility device to walk),¹ the ability to sustain upright standing balance has a profound effect on activities of daily living and transitions between positions.^{2,3} Decreased static standing balance control has been correlated to less functional abilities of the child as well as increased caretaking.² Static and dynamic postural control in standing are achieved and sustained by actively aligning body segments, maintaining body position under the influences of the environment and gravity, and stabilizing center of mass (COM) within the base of support.^{4,5} The capacity to maintain standing in children with CP is challenged by impaired neuromuscular control, as evidenced by increased postural sway, coactivation of muscles around a joint, temporal-spatial disorganization with increased proximal to distal activation, decreased muscular strength, and longer latency in response to balance threats.^{4,6,7}

In addition to neuromuscular impairments, biomechanical alignment impairments contribute to deficits in postural control. Erect standing in the sagittal plane in typical development is characterized by the gravitational force vector being anterior to the ankle joint, through or slightly anterior to the knee joint, through or slightly posterior to the hip joint and with a predictable relationship to normal spinal curves. This alignment is the most energy efficient and provides a basis for upright stability.⁸ Muscle weakness and contractures limit the acquisition of erect standing and place increased energy demands on children with CP. Children with CP may have disturbances of sensation and processing that limit interaction with the environment, such that perception may become inaccurate, negatively influencing the resulting action.^{9,10}

Endurance capacity is impaired for children with CP, GMFCS levels II and III, who expend 1.4 times greater energy during standing than children developing typically.¹¹ Verschuren et al¹² found that children with GMFCS level III had increased muscle activity and energy expenditure during standing, meeting the standard for “light activity,” and Rose et al¹³ demonstrated that walking required 2 to 3 times as much energy for children with bilateral CP compared with peers developing typically.

Muscle recruitment patterns during standing postural responses of children with CP GMFCS levels I to III have a stronger tendency toward proximal to distal rather than distal to proximal recruitment seen in children developing typically.^{3,14,15} Atypical recruitment patterns and higher rates of coactivation of agonist and antagonist muscles experienced by children with CP result in inefficient and disorganized postural control.^{3,14} They work harder with less efficient ability to maintain balance than their peers developing typically.

Bernstein’s motor control theory posits that the central nervous system interacts with redundant biomechanical possibilities for a particular movement by devising efficient patterns for function.¹⁶ In learning a new movement, typical learners gain control by freezing degrees of freedom (DOF) (consider the first stiff steps of an infant learning to walk). Over time, the learner releases DOF to explore and expand the flexibility of movement to build toward an ultimate goal of coordinated and efficient movement. Dynamic

systems theory extends Bernstein's work and places emphasis on the interaction and self-organization of multiple subsystems within the main determinants of movement: the person, the task, and the environment.¹⁷ The refinement of kinetic and kinematic responses that underlie balance control emerges, as children interact with their environment, experiment with movement, and discover their own solutions.¹⁴ Children with CP are challenged to modulate and fine tune movement for balance control. Carlberg and Hadders-Algra¹⁸ suggest that restricting DOF might be an effective therapeutic approach to learning postural control for functional tasks.

This study was born from a consistent clinical observation that distal stabilization, in the form of ski boots, provided benefit for training standing postural control in children with CP. Children stood for a longer duration with more erect postural alignment. By limiting DOF at the ankle and foot, children with CP explored DOF proximally, at the trunk, hip, and knee, anecdotally leading to more effective balance responses.

Few researchers have investigated the effects of different orthotics on standing balance. Rha et al¹⁹ assessed 21 children with bilateral spastic CP (age 6 ± 1.09 years) who could stand independently for 30 seconds. Their study found no improvement in standing postural stability with the use of hinged orthoses compared with bare feet (BF). Wesdock and Edge²⁰ demonstrated increased standing duration in 11 children with GMFCS levels II, III, and IV (ages 4–13 years) while wearing solid ankle-foot orthoses (AFOs) with wedging compared with BF and non-wedged orthoses. Bahramizadeh et al²¹ assessed 8 children with spastic diplegia, GMFCS levels I and II (age 8 ± 2 years) and reported the effects of floor reaction AFOs on standing postural control and found improved knee extension but no significant changes in center of pressure (COP) measurements.

Ankle-foot orthotic footwear combination (AFOFC) strives toward normal ground reaction forces and boney alignment via individualized prescription of the angle of the ankle of the AFO (AAAFO) and optimizing the shank-to-vertical alignment (SVA) for stability in static standing and mid-stance of gait (knee aligned over mid-foot).²² The tuning of the AAAFO is the product of a published algorithm, which considers the length, stiffness, and strength of the gastrocnemius-soleus muscles and integrity of the transverse arch.²² The inclined position of the shank lowers the stance limb and vertical excursion of the COM, which may contribute to stability and energy conservation,²³ facilitating optimal kinematics to reduce muscle demand for children with CP.²⁴ Assessment of postural control and muscle activity in standing with the AFOFC or its component parts (AAAFO, SVA) has not been investigated.

Few studies have examined muscular activity via electromyography (EMG) during quiet standing in children with CP. Burtner et al²⁵ studied 4 children with spastic CP and at least 1 year of independent walking (ages 3½–15 years) and found no changes to muscle recruitment or activation patterns in response to perturbation when comparing BF, hinged, and solid AFO, but reported that solid AFO was associated with decreased gastrocnemius firing. Radtka et al²⁶ found that 12 children with spastic diplegic CP (average age 7.5 years) had increased coactivation and prolonged shank and thigh EMG activity during the stance phase of gait both with and without orthotics when compared with children developing

typically. Rosenberg and Steele²⁴ suggested that appropriate tuning of AFO properties may reduce muscle demand for children with CP.

The purpose of this exploratory study was to investigate whether distal control orthoses (DCOs) with individually optimized AAAFO and SVA, and controlling of distal DOF, improved standing in participants with CP, compared with BF and the participant's AFO. First, we hypothesized that DCO would improve standing duration, decrease center of gravity (COG) sway velocity, and provide more stable medial-lateral (M-L) and anterior-posterior (A-P) alignment. Second, we hypothesized that participants wearing DCO compared with BF or AFO would demonstrate: (1) decreased overall EMG activity indicating increased standing efficiency and (2) decreased proximal relative to distal muscle activity, which is more consistent with children developing typically.

METHODS

Participants

A convenience sample of 10 participants with CP was selected for this study (Table 1). The inclusion criteria were: (1) GMFCS level III¹; (2) age 7 to 20 years; (3) no fixed contractures beyond passive range of motion (ROM) values described by Burtner et al²⁵ (<16° hip flexion; 10° knee flexion, or 10° ankle plantar flexion); (4) normal or corrected functional vision; (5) no other medical diagnosis affecting gross motor skills; (6) capacity to follow simple commands; and (7) ability to stand independently for 10 seconds while BF. The exclusion criteria included: (1) scoliosis greater than 25°, (2) hemiplegia, (3) orthopedic or neurosurgery within 6 months of recruitment, and (4) botulinum toxin within 8 weeks of casting for orthoses or within 3 months of testing phase of the study.

This exploratory within-subject study was approved by the Institutional Review Boards of the National Institute of Health (NIH) and the Rocky Mountain University of Health Professions. Guardian informed consent and assent forms were signed by parents and participants younger than 18 years, respectively. For participants older than 18 years capable of making independent decisions, adult consent forms were signed.

Equipment

Participants were casted and fitted with custom DCO with an anterior shell by an experienced pediatric orthotist. The orthotic design limited ankle-foot movement and provided A-P and M-L anatomical support. The AAAFO was determined by the stiffness of the gastrocnemius-soleus muscle and available length on first resistance, as measured in subtalar neutral with knee extended. The orthoses were posted to a vertical shank and wedged to optimize SVA toward normal values (10°–12°). More erect standing (less reclined thigh) was demonstrated using clinical observation of sagittal alignment with a posture grid (Figures 1A and 1B). The SVA was measured with an inclinometer on anterior shank of orthotic (range 10°–14°). Participants wore flat-sole shoes for AFO and DCO conditions (Figure 1C).

The Neurocom Balance Master System (long force plate) was used to assess postural control, eyes open (EO) and eyes closed (EC), with the firm ground conditions of the

modified Clinical Test of Sensory Interaction on Balance (mCTSIB). These tests are considered gold standards for balance testing of children.²⁷

Surface EMG data were collected during the mCTSIB using Trigno wireless EMG system (Delsys Inc, Natick, Massachusetts). An opening of 4 cm × 6 cm was created on DCO orthoses over muscle bellies of the anterior tibialis and proximal head of the medial gastrocnemius for electrode placement.

Procedures

Each participant had a history and physical examination by an NIH physician and was cleared for participation in the study. Participants were tested in 3 ankle-foot conditions in this order: DCO, BF, and current AFO (Table 1). This testing order was determined to be least disruptive to wireless EMG leads, which, if offset, might compromise validity of EMG comparison across conditions. Testing procedures included standing duration tests on a level surface, followed by mCTSIB. Rest was provided between tests as determined by each child's expressed and visible level of fatigue.

For standing duration tests, participants stood with feet side by side and then step stance, with self-selected foot position. Participants were instructed to stand independently for up to 1 minute in each posture and were timed with a hand-held stopwatch to the nearest second.

The mCTSIB protocol measured COG sway velocity, M-L and A-P alignment, and loss of balance (LOB). Participants performed 1 to 2 instructional trials and 3 data collection trials; mean scores were used for data analysis. They were closely supervised and positioned with a walker in front and a bench behind them as safeguards. For the mCTSIB protocol, the goal was standing first with EO, then EC.

EMG data were collected from 7 muscles bilaterally, in accordance with SENIAM recommendations,²⁸ via wireless surface electrodes on anterior deltoid, low lumbar erector spinae, rectus abdominis, rectus femoris, semitendinosus, gastrocnemius, and anterior tibialis. EMG data were filtered using a 5-Hz pass filter, full-wave rectified, then a second-order low-pass Butterworth filter with a cutoff frequency of 2 Hz to create a linear envelope with Visual 3D software. During the mCTSIB, EMG recordings (Nexus [Vicon Motion Systems, Denver, Colorado]) were synchronized with Neurocom data collection and video data were recorded. To allow for comparison of EMG activation across conditions, the area under the curve of the linear envelope (a function of amplitude and time) was calculated for each muscle in each trial.²⁹ Total EMG is the sum of the areas under the curve for all muscles within a trial, which was done to compare the overall muscle activity during the different ankle-foot conditions.

Data Analysis

Data were analyzed for normal distribution for the 12 dependent variables. A-P/M-L alignment and EMG data were normally distributed, but not correlated. Consequently, repeated-measures analysis of variance (ANOVA), rather than multivariate analysis of variance (MANOVA), was used for comparison across ankle-foot conditions for these variables. The ratio of anterior deltoid/gastrocnemius EMG activity, to represent proximal/

distal relationships, was assessed with repeated-measures ANOVA across the 3 ankle-foot conditions. A mixed 2-way repeated-measure ANOVA examined the interaction and effect of time on learning for the within subject-repeated measures. COG sway velocity, LOB, and standing duration were not normally distributed. Friedman 2-way ANOVA by ranks was used for repeated measures and to determine effect size. Effect size interpretation was based on published guidelines.³⁰

Because of the nature of this preliminary study (to avoid type II error), post hoc pairwise comparisons were conducted using least significant difference method (equivalent to no adjustments) following statistically significant ANOVAs. Likewise, no adjustments to multiplicity were made when performing post hoc Wilcoxon signed-ranks tests following significant results with the Friedman test. The Statistical Package for the Social Sciences (SPSS) 26.0 was used for analyses with a significance level of $P = .05$.

RESULTS

Standing Duration

Side-by-side stance duration was statistically significant across ankle-foot conditions with pairwise comparison with significantly longer duration in DCO versus BF. Effect sizes were large for DCO versus BF, and DCO versus AFO. Standing duration in step stance was not statistically significant, but effect sizes were large DCO versus BF and DCO versus AFO (Table 2).

Center of Gravity Sway Velocity (mCTSIB)

COG sway velocity when wearing DCO in EO and EC conditions was significantly decreased compared with BF and AFO, with large effect sizes in each comparison. No statistically significant differences were found in BF versus AFO conditions in EC or EO (Table 2).

Loss of Balance (mCTSIB)

With both EO and EC, statistically significant differences were recorded in LOB among ankle-foot conditions. In EC pairwise comparisons, less LOB occurred in DCO versus BF and AFO versus BF. Effect sizes were large for all EC comparisons and for EO comparisons of DCO versus BF and AFO versus BF (Table 2).

M-L and A-P Alignment (mCTSIB)

No differences were noted in M-L alignment among ankle-foot conditions for EO or EC. The A-P alignment showed statistically significant increased anterior translation of COP in DCO compared with BF and to AFO (EO and EC) with large effect sizes. No differences were found in A-P alignment between BF and AFO conditions (Table 3).

Measures of EMG

Differences in EMG activity with EO led to post hoc tests demonstrating statistically significant decreased muscle activity with AFO versus BF. The DCO versus BF comparison

trended toward statistically significant decreased activity. No statistically significant EMG differences were detected among ankle-foot conditions with EC (Table 3).

Testing Order

Mixed 2-way ANOVA (time $\times$ condition) showed no interaction effects for A-P alignment, suggesting order of testing did not affect outcome either positively (motor learning) or negatively (fatigue).

Photographs and Video Observations

Photographs portray improved standing postural alignment comparing BF, AFO, and DCO (Figures 2A, 2B, and 2C). Video review of participants during mCTSIB revealed marked movement oscillations, notably of the arms, trunk, and pelvis, in the BF condition. In AFO, participants were steadier but often asymmetrical, with less pronounced oscillations. Arm, trunk, and pelvic movements were the least in DCO compared with other conditions, and posture was generally more symmetrical. Participants appeared steadier with less observed effort to sustain standing (see the Supplemental Digital Content Video, available at: <http://links.lww.com/PPT/A321>). In DCO, participants had a statistically significantly decreased COG sway velocity, compared with the other conditions, which may objectively lend support to these clinical observations.

DISCUSSION

This study evolved from clinical observations of longer duration and more erect standing when stabilizing the ankle-foot complex and shank of the child with CP GMFCS level III. The intent was to compare the influence of 3 ankle-foot conditions on standing postural control and muscle activity. Participants in DCO versus BF had a decrease in COG sway velocity (EO and EC) and LOB (EC), and increase in side-by-side standing duration with large effect sizes, supporting our clinical findings. COG sway velocity (EO and EC) was significantly less in DCO versus AFO with large effect size, but not different in AFO versus BF. The lack of difference between AFO and BF may prompt reservations about the effectiveness of AFOs for improving this measure of stability.

The forward translation of A-P alignment for the child wearing DCO provides an advantage to the kinematic alignment in the sagittal plane. Children with CP often present with excessive or insufficient dorsiflexion and/or plantar flexion initiating a cascade of segmental challenges in more proximal joints. By eliminating grossly abnormal shank alignment, the DCO facilitates more successful proximal strategies up the kinematic chain for standing. With the ankle, foot, and shank stabilized by the DCO, children may have the opportunity to gradually master control of the knee, hip, and trunk under the new condition of practice.^{31,32} In the dynamic systems motor control model, the concept of depth of “attractor well” impacts potential for change in motor strategies such that a deep well is an entrenched movement unlikely to change. Perhaps use of DCO in balance training can function to facilitate a self-organizing process through the ankle-foot constraints, effectively making the “attractor well” for poorly functioning standing more shallow, and thereby more susceptible to change.¹⁷ Additionally, the freezing of distal DOF, as described by Bernstein, may afford

the exploration of motor organization more proximally at the knees, hips, and trunk. As stated by Pavão et al,³ children must be able to coordinate their joint segments, generate the proper sequence of muscle contractions, and keep their COM within the base of support to explore the world and change their positions voluntarily.

Decreased muscle activation with a solid AFO has been reported with improvement in endurance during standing and/or walking.^{24,33,34} Decrease in EMG activity may reflect any or all of the following: less agonist effort, more efficient timing of muscle activity, or less coactivation of agonist and antagonist. The expectation of decreased EMG activity in DCO compared with AFO was not realized. This finding may be due to the novel experience of wearing the DCO, which may increase overall muscle activity,³⁵ compared with the daily familiarity participants had with their current AFO.

We hypothesized that DCO would impact total muscular activity and proximal-distal muscular relationships to improve static standing, but our findings did not corroborate this. Like Burtner et al,²⁵ we found no difference in proximal/distal activity. While the DCO decreased COG sway velocity, the EMG data do not support any associated changes in muscular activity in DCO. Zaino and McCoy³⁶ proposed that COG is a more constant control parameter during standing compared with the more variable neuromuscular system. The association of the biomechanical alignment offered by the DCO with less evidence of EMG influence is consistent with the Zaino and McCoy proposal.

Physical therapists, recognizing the influence of multiple systems on balance, are challenged to provide evidence-based interventions to improve postural control. This study does not establish which component(s) of our intervention was/were most influential in improving standing control: restriction in ankle ROM, facilitation of improved shank alignment, or both. Our findings provide preliminary evidence that DCO may provide a useful therapeutic tool for postural control training.

Future Considerations

Future research may include the study of independent contributions of ankle-foot alignment and the influence on postural control. Instrumentation of kinematics may further support the benefits of DCO. A period of training in the DCO may be necessary to demonstrate change in muscular activity and begs further questions: (1) Would training in DCO continue to improve postural control outcomes over time? (2) If so, would releasing the ankle-foot DOF gradually be an effective therapeutic strategy to improve postural control? (3) Can we assure that motion restriction at the ankle with DCO has no negative effects?

Limitations

This exploratory study has several limitations including small sample size assessing 12 dependent variables (limiting power of the study), wide age range with heterogeneity of developmental and physiological maturity, and nonuniformity in prescribed orthoses. While DCO significantly impacted COG sway velocity, LOB (EC), and standing duration, generalization of these findings is interpreted with caution, as it is important to recognize that each child with CP requires judicious clinical intervention toward their unique responses.³⁷

CONCLUSION

This study provided evidence that use of DCO, a biomechanical substitute for an unstable ankle-foot, decreased COG sway velocity, decreased LOB, and increased standing duration in children with CP. DCO may provide a first step in therapy to improve standing, allowing exploration of DOF more proximally, and time and opportunity to practice coordination of intersegmental joints for more effective postural control. While EMG data revealed no differences in muscular effort among the 3 ankle-foot conditions, possible gains in standing stability in the DCO may, with practice, afford a reduction in muscular effort during standing. Given the relationship between static postural control and functional abilities in children with CP, further evidence-based research is essential for identifying strategies to improve standing postural control.³

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.

(A) Participant standing barefoot. (B) Standing more erect with distal control orthoses at posture grid. (C) Distal control orthoses. This figure is available in color online (www.pedpt.com).

Fig. 2. Participant standing alignment with 3 ankle-foot conditions of: (A) barefoot, (B) ankle-foot orthoses, and (C) distal control orthoses. This figure is available in color online (www.pedpt.com).

TABLE 1

Description of Participant Demographics and Individually Prescribed Ankle-Foot Orthoses (Orthoses Worn Daily by Participants) and the Distal Control Orthoses Fabricated With Ankle-Foot Range Restriction With SVA Set to Gain Optimal Alignment

Subject	Sex	Age	Walking Device	Prescribed AFO Style	DCO AA	DCO SVA
1	F	15	Posterior walker	SMO	Neutral	12
2	F	16	Posterior walker	SMO	Neutral	11
3	M	17	Canes	SMO	3° PF posted to neutral	10
4	M	7	Posterior walker	Articulated AFO; free DF; PF stop	2° PF posted to neutral	14
5	M	10	Posterior walker	SMO	2° PF posted to neutral	10
6	M	9	Walking sticks	SMO	0° DF	12
7	F	20	Lofstrand crutches	Solid AFO at 0° DF	0° DF	10
8	F	10	Posterior walker	Articulated AFO with check strap; limited DF; PF stop	3° PF posted to neutral	12
9	M	11	Posterior walker	SMO with DF assist orthotic on left	2° PF posted to neutral	13
10	M	18	Canes	Articulated AFO; free DF; PF stop	3° PF posted to neutral	11

Abbreviations: AA, angle of ankle; AFO, ankle-foot orthoses; DCO, distal control orthoses; DF, dorsiflexion; F, female; M, male; PF, plantar flexion; SMO, supramalleolar orthoses; SVA, Shank-to-vertical angle.

TABLE 2

Friedman 2-Way ANOVA by Ranks of Ankle-Foot Conditions on COG Sway Velocity, LOB, and Standing Duration^a

	COG Sway Velocity	EO	COG Sway Velocity	EC	LOB	EO	LOB	EC	Standing Duration	Side × Side	Standing Duration	Step Stance
Friedman 2-way ANOVA by ranks												
Omnibus significance	$P = .001^b$		$P = .001^b$		$P = .05^b$		$P = .004^b$		$P = .029^b$		$P = .054$	
Wilcoxon												
DCO vs BF	$P = .008^b$		$P = .008^b$		$P = .109$		$P = .028^b$		$P = .043^b$		$P = .043$	
DCO vs BF effect size	0.84		0.89		0.51		0.73		0.64		0.64	
DCO vs AFO	$P = .007^b$		$P = .012^b$		$P = 1.00$		$P = .068$		$P = .109$		$P = .046$	
DCO vs AFO effect size	0.85		0.84		0.32		0.60		0.51		0.63	
BF vs AFO	$P = .235$		$P = .141$		$P = .109$		$P = .028^b$		$P = .5$		$P = .753$	
BF vs AFO effect size	0.38		0.49		0.51		0.73		0.21		0.10	

Abbreviations: AFO, ankle-foot orthoses; ANOVA, analysis of variance; BF, barefoot; COG, center of gravity; DCO, distal control orthoses; EC, eyes closed; EO, eyes open; LOB, loss of balance.

^aFriedman effect size interpretation³⁰: small (0.10), medium (0.3), and large (0.5).^bSignificance at $P < .05$.

TABLE 3

Repeated-Measures ANOVA of Ankle-Foot Conditions on M-L and A-P Alignment and Total EMG^a

	A-P Alignment EO	A-P Alignment EC	M-L Alignment EO	M-L Alignment EC	EMG EC	EMG EO
Omnibus significance	$P = .007^b$	$P = .013^b$	$P = .518$	$P = .628$	$P = .386$	$P = .018^b$
Pairwise comparison						
DCO vs BF	$P = .011^b$	$P = .018^b$	$P = .389$	$P = .665$	$P = .423$	$P = .053$
DCO vs AFO	$P = .041^b$	$P = .050^b$	$P = .455$	$P = .329$	$P = .957$	$P = .181$
BF vs AFO	$P = .408$	$P = .549$	$P = .389$	$P = .843$	$P = .269$	$P = .008^b$
Effect size	0.481	0.417	0.090	0.039	0.113	0.441
Power	0.729	0.792	0.126	0.083	0.135	0.732

Abbreviations: AFO, ankle-foot orthoses; ANOVA, analysis of variance; A-P, anterior-posterior; BF, barefoot; DCO, distal control orthoses; EC, eyes closed; EMG, electromyography; EO, eyes open; M-L, medial-lateral.

^aPartial η effect size interpretation³⁰: small (0.01), medium (0.09), and large (0.25).

^bSignificance at $P < .05$.