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Authors: S.M. Steinl, P.J. Sparto, C.G. Atkeson, M.S. Redfern, L. Johannsen

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Interpersonal interactions for haptic guidance during balance exercises

S. M. Steinl[a], P.J. Sparto[b], C.G. Atkeson[c], M.S. Redfern[d], L. Johannsen[e]

[a]*Department of Sport and Health Science, Technical University Munich, Munich, Germany*

[b]*Department of Physical Therapy, School of Health and Rehabilitation Sciences, University of Pittsburgh, Pittsburgh, USA*

[c]*Robotics Institute and Human-Computer Interaction Institute, Carnegie Mellon University, Pittsburgh, USA*

[d]*Department of Bioengineering, Swanson School of Engineering, University of Pittsburgh, Pittsburgh, USA*

[e]*School of Health Sciences, University of East Anglia, Norwich, UK*

Corresponding Author:

S.M. Steinl, Department of Sport and Health Science, Technical University Munich, Munich, Germany, Georg Brauchle Ring 60/62, 80992 Munich

saskia.steinl@tum.de

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Study Highlights

- passive support mode demonstrated its advantages in increased strength of the IPC
- active support mode decreased the postural sway to a greater extent.
- more partnership based methods should be considered for balance rehabilitation
- postural control can be responsive to social factors

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Abstract

Background

Caregiver–patient interaction relies on interpersonal coordination during support provided by a therapist to a patient with impaired control of body balance.

Research question

The purpose of this study was to investigate in a therapeutic context active and passive participant involvement during interpersonal support in balancing tasks of increasing sensorimotor difficulty.

Methods

Ten older adults stood in semi-tandem stance and received support from a physical therapist (PT) in two support conditions: 1) physical support provided by the PT to the participant's back via an instrumented handle affixed to a harness worn by the participant ("passive" interpersonal touch; IPT) or 2) support by PT and participant jointly holding a handle instrumented with a force-torque transducer while facing each other ("active" IPT). The postural stability of both support conditions was measured using the root-mean-square (RMS) of the Centre-of-Pressure velocity (RMS dCOP) in the antero-posterior (AP) and medio-lateral (ML) directions. Interpersonal postural coordination (IPC) was characterized in terms of cross-correlations between both individuals' sway fluctuations as well as the measured interaction forces.

Results

Active involvement of the participant decreased the participant's postural variability to a greater extent, especially under challenging stance conditions, than receiving support passively. In the passive support condition, however, stronger in-phase IPC between both partners was observed in the antero-posterior direction, possibly caused by a more critical (visual or tactile) observation of participants' body sway dynamics by the therapist. In-phase cross-correlation time lags indicated that the therapist tended to respond to participants' body sway fluctuations in a reactive follower mode, which could indicate visual dominance affecting the therapist during the provision of haptic support.

Significance

Our paradigm implies that in balance rehabilitation more partnership-based methods promote greater postural steadiness. The implications of this finding with regard to motor learning and rehabilitation need to be investigated.

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Keywords: Interpersonal coordination, balance rehabilitation, social postural coordination, haptic support

1. Introduction

Falls and fall related injuries in older adults are a public health issue [1, 2]. Balance exercises, however may reduce falls risk [3]. In balance rehabilitation, a physical therapist (PT) manipulates the provision of sensory cues during sensorimotor training to facilitate motor learning, and control of body balance [4-6].

The factors governing sensorimotor interactions between therapist and client, however are poorly understood [7]. Interpersonal sensorimotor interaction can be classified into cooperation and collaboration [8]. In contrast to collaborative interactions that do not integrate a priori role assignments, roles are assigned a priori to each participant in cooperative interactions. For example during balance exercises, this can lead to an allocation of sub-tasks, such as provision of haptic balance support by a therapist and reception by the client involved in the balancing task [9].

Additional tactile feedback is a reliable approach to augment control of body balance [10]. In the traditional paradigm ("active" light touch), a participant is controlling the upper limb directly, which is contacting the external haptic reference [11]. Hereby, the movement degrees of freedom of the contacting limb are used for precision control of the contact force with the control of body sway as a separate process [12]. In addition to the haptic feedback signal, the output of fingertip control could serve as a signal to control sway [13]. In non-manual, "passive" light touch, the contact is delivered to a participant's body segment. A participant is less able, to control the precision by which the contacting force is applied [13]. Here, the movement degrees of freedom available to a participant for controlling the contact force are limited by the current postural degrees of freedom, thereby creating a direct equivalence between control of body sway and precision of the contact.

Passive light touch with an earth-fixed reference results in proportional sway reductions in the range of 20%-30% [13]. This is similar to what has been reported in studies involving fingertip light touch [i.e. 14]. Interpersonal fingertip touch (IPT) leads to lesser sway reductions of around 9-15% [9, 14-17]. The reason for this diminished effect could lie in the fact that the contact reference is not earth-fixed but shows own motion dynamics, which might make dis-

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ambiguation of the haptic signal in terms of own sway-related feedback more challenging. Johannsen et al. [9] assessed “passive” IPT in neurological patients as well as chronic stroke and reported sway reductions between 15%-26%. In stroke patients, passive, trunk-based IPT [9], nevertheless, seemed more beneficial than fingertip IPT [16].

In our study, we directly contrasted the effects of active and passive support modes on body sway in a therapeutic setting. We measured the interaction forces between a physiotherapist and participants and characterized the interpersonal postural coordination (IPC) between both partners. We predicted that the participant would demonstrate the greatest sway reductions when passive IPT was provided to the trunk with no involvement in contact precision control. We increased the sensory challenges imposed by the balance task (foam surface, eyes closed, pitch head movement) and assumed that with increasing difficulty, the benefit of IPT would increase as well potentially in interaction with the specific IPT mode.

2. Methods

Participants

Ten older adults without significant neurological or orthopedic history, between the age of 71 and 86 years (mean age 79 yrs, SD= 5; 5 females, 5 males; all right-handed for writing) participated in this study. One PT (16years of experience) provided support.

Recruitment and Exclusion criteria

Participants were recruited from a sample of screened healthy elderly subjects from a preliminary study [18]. This study was approved by the Institutional Review Board of the University of Pittsburgh.

Demographic data

Participants completed the Activities-specific Balance Confidence Scale (ABC) questionnaire [19] and the Functional Gait Assessment [20] prior to the experiment. The participants reported a balance confidence level between 74% and 100% (mean 94%, SD=8). The Functional Gait Assessment (FGA) is a modification of the Dynamic Gait Index (DGI) that uses higher level gait tasks [20]. Participants achieved scores between 17 and 30 in the FGA (mean 26, SD=5).

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Experimental Design

Participants performed 2 sets of 6 randomized balance exercises during two different conditions: passive support (PS) and active support (AS) (Figure 1). In the PS condition, the PT who was in bipedal stance with full vision, stood behind the participant and lightly held on to an instrumented handle mounted on the back of the participant's vest and applied stronger support only when he felt the participant required firmer assistance to maintain upright balance. In the AS condition, the PT and the participant faced one another and simultaneously held on to the handle. Participants were instructed to stand as stable as possible with their arms crossed in front of their waist (PS) or to stand as stable as possible while holding on to a handle (AS). For each set of six balance conditions participants completed a partial factorial design of the conditions (see Fig 1D). These exercises were chosen across a range of difficulty based on a preliminary study [18].

Instrumentation

The participant and PT stood on separate force platforms (Bertec, Columbus, Ohio, USA) that measured ground reaction forces and moments at a sampling rate of 120 Hz (see Fig 1A, B). A tri-axial load cell (DSA-03A TecGihan, Japan) was mounted to a custom-made handle and bracket which was secured to the back of a support vest worn by the participant to measure forces during the PS condition (see Fig 1A). Force plate and load cell data were collected by the same data acquisition system (National Instruments, Austin, TX). During the AS condition, the handle was removed from the vest and a second handle was attached to the bracket for the participant's use (see Fig 1B).

Procedure

Participants stood in semi-tandem stance by placing their feet so that the medial borders were touching, and moving their dominant foot backward by a half of foot length [21]. During the foam surface conditions, participants stood on foam (AIREX Balance Pad S34-55, height 6cm, length 51 cm, width 40 cm). During the pitch condition, participants moved their head over a total range of 30 degrees at 1 Hz by following a metronome [22]. Trials lasted 30 seconds and participants wore a safety harness.

Data reduction and statistical analysis

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The force platform and load cell data were transformed into center of pressure (COP) and handle force measurements, respectively, using calibration equations The antero-posterior (AP) and medio-lateral (ML) components of the COP and the AP component of the handle force were extracted. All data time series were smoothed using a dual-pass, 4th order Butterworth lowpass filter (cutoff=10Hz). COP data were numerically differentiated to produce COP velocity measures. Velocity information is the predominant source of body sway control [23] therefore the root-mean-square of the AP and ML COP velocity (RMS dCOP) were the primary postural control measures. The IPC was estimated by computing the cross-correlation functions between both participants' COP velocity time series.

Cross-correlations were computed within a range of minimum and maximum time lags between -/+3 s. We used the standard MATLAB cross correlation function which measures the dependence between two signals [24, 25]. The largest maximum (in-phase behavior) and minimum (anti-phase behavior) cross-correlation coefficients and corresponding time lags were extracted. The cross-correlation coefficients were Fisher Z-transformed for statistical analysis.

SPSS version 23 was used for statistical analysis. A linear mixed model analysis with support mode (2 levels: active and passive) and condition (6 balance exercises) effect as well as the support * condition interaction was performed. For the estimation of the model we used a maximum likelihood method. Postural sway parameters (RMS) were analyzed including subject as a random effect while IPC parameters (correlation coefficients, lags) and forces were analyzed using only fixed effects. A diagonal covariance structure was used for repeated effects in the mixed model [26]. An alpha level of 0.05 was used for level of significance, and post-hoc comparisons were computed using Sidak adjustment.

3. Results

3.1. Postural control

Sway velocity in AP direction

Significant support ($F(1,58.5) = 22.8$, $p < 0.001$) and condition ($F(5,28.5)=80.6$, $p<0.001$) effects were found for participant RMS dCOP in the AP direction (Figure 2). The passive support led to higher sway velocity production. The sensory conditions generated progressively increased sway velocity (see Fig. 2A).

Sway velocity in ML direction

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Analysis of the RMS dCOP in the ML direction generated similar support ($F(1,57.5)=51.3$, $p<0.001$) and condition ($F(5,25.9)=59.2$, $p<0.001$) effects as in the AP direction, but there was also a significant interaction between condition and support ($F(5,25.8)=3.90$, $p=0.001$) (Figure 2B). The interaction indicates that there was greater difference in the amount of sway velocity between passive and active support conditions as the balance conditions became more challenging. The difference in sway velocity ranged from approximately 18.5 mm/s in the firm surface, eyes open, head still condition to 58 mm/s during the foam surface, eyes closed, head pitch condition.

3.2 Handle Forces

Average AP handle force

A significant effect of support mode ($F(1,46.2)=8.22$, $p=0.01$) on the average handle force was found (Figure 3A). A mean force of 1.7 N (SD 0.5 N) in the posterior direction on the handle was observed during the passive support trials. During the active support trials, the forces of the PT and participants counteracted one another on average, with a mean force of 0.01 N (SD=0.05 N) towards the PT. A significant effect of sensory condition ($F(5,22.2)=4.0$, $p=0.01$) was found. Larger posterior forces on the handle were exerted during the foam, eyes closed, and passive support conditions compared with much smaller force exertion during the other conditions. During the active support trials, a pattern emerged in which the force was directed toward the participant in the easier conditions, and toward the PT in the foam, eyes closed conditions. Lateral forces were also minimal (see Fig. 3A).

Variation in AP handle force

The magnitudes of variation of handle forces applied between the PT and participant, as measured by the standard deviation of the time series, are shown in Figure 3B. A progressive increase in variation in forces occurred as the sensory conditions became more difficult ($F(5, 21.8)=18.4$, $p< 0.001$).

3.3 Interpersonal coordination of postural sway

Minimum cross correlation coefficients between participant and PT

Figure 4 displays the minimum (i.e. anti-phase) cross correlation coefficients between the COP velocity of the PT and participant. A significant condition effect was found in both the AP (Figure 4A, $F(5,29.7)=9.2$, $p<0.001$) and ML directions (Figure 4B, $F(5, 37.1)=3.9$, $p=0.01$). In the AP direction, IPC anti-phase behavior was larger in the eyes closed condi-

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tions. In the ML direction, there was less anti-phase IPC in the firm surface, eyes open, head still condition.

Maximum cross correlation coefficients between participant and PT

The maximum (i.e. positive) cross-correlations were greater in absolute magnitude than the minimum (negative) cross-correlations, indicating that the in-phase IPC was more prominent than the anti-phase IPC. The IPC in-phase behavior of the COP velocity in the AP direction demonstrated significant support, condition and interaction effects (Figure 4C). Lower average interpersonal cross-correlation coefficients were found in AS 0.28 (SD 0.02) than in PS 0.34 (SD 0.02) in the AP direction ($F(2,101.8)=13.4$, $p<0.001$), which indicated greater strength of the in-phase IPC in the passive mode. The sensory conditions differed ($F(5,34.2)=20.8$ $p<0.001$), which showed increasing IPC during the more difficult sensory conditions, similar to the pattern of results of the RMS dCOP. A significant interaction between support and exercise mode ($F(5,34.2)=2.7$ $p=0.04$) demonstrated greater IPC during the active support mode for the firm surface, eyes open, head still condition, in contrast with greater IPC during the passive support mode for all other conditions. The in-phase coordination in the ML directions showed a significant condition effect only ($F(5,35.8)=14.24$ $p<0.001$).

3.4 Time lags in IPC between participant and PT

We found a significant support mode effect ($F(1,90.6)=6.6$, $p=0.02$; passive mean = -287 ms SD = 13 ms; active mean = 210 ms SD = 13 ms) (Figure 5A; anti-phase IPC). The PT led in all but the third sensory condition (AS) and followed in all but the second and third sensory conditions (PS). Figure 5C (in-phase IPC) demonstrates a pattern in which the PT was always the follower (AS: mean= 159 ms SD=17 ms; PS: mean= 323 ms SD=21 ms) with the exception of the easiest sensory condition (firm, EO, still) in active mode.

4. Discussion

We aimed to contrast the effects of two different modes of client participation in the provision of interpersonal light touch balance support by a therapist to balance-challenged older adults.

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4.1. Postural control

In both directions, the active support mode resulted in less participant sway velocity compared with the passive support mode. Proportional sway velocity difference between both modes was 32% of passive condition, which is similar to passive LT sway reductions with an earth-fixed reference or fingertip LT [14, 27]. An interaction between support mode and sensory condition for sway in the ML direction indicated that the active support mode provided a greater benefit with greater sensory disruption. The observation that more active participation in the control of contact force precision resulted in reduced sway under conditions of greater sensorimotor destabilization was unexpected as in previous studies the comparative proportional benefit of passive trunk-based IPT on body sway tended to be greater than IPT at the fingertips.

The difference between the two IPT modes in this study could rest on stronger and less ambiguous haptic feedback from the grasp of the handle or processes of anticipatory postural control and voluntary force precision control in the active IPT mode. Wing et al. [28] investigated the coupling between grip force during one-handed precision grasp on a manipulandum and concurrent postural adjustments in anticipation of dynamic and static loads during horizontal pulling and pushing. They demonstrated a functional linkage between grip force adjustments anticipating changes in load force on the manipulandum and ground reaction torque in anticipation of self-imposed balance perturbations due to the pushing and pulling motion. They suggested that an efferent signal controlling grip force could facilitate the prediction of upcoming postural load and appropriate postural adjustments [28]. Further, minimization of the interaction force and its variability could have resembled the goal of a so called “suprapostural” task resulting in proactive, task-adapted body sway reductions [29, 30]. As the latter mechanism might apply to fingertip IPT too, we speculate that an efferent grip force control signal contributing to anticipatory postural control facilitated postural stability primarily in this study instead.

By facing the participant in active mode, the therapist might have received clearer social cues about postural destabilization of the participant that facilitated internal simulation of a participant’s sway dynamics for the anticipation of instabilities and need for support [31]. For example, the sight of another person can improve an individual’s ability to compensate for imbalance [32].

4.2 Handle Forces

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It needs to be considered that in the passive IPT mode, strength of the contacting force was upregulated intermittently based on the therapist's visuotactile assessment of a participant's current state of postural stability. In the easier sensory conditions, the interaction forces remained relatively low, which possibly indicates the relative absence of active stabilization of participants' sway by the therapist. The interaction forces fell into the range from 4 N to 6 N in the two most challenging conditions (foam surface), which could imply more continuous in addition to stronger haptic support.

Nevertheless the stronger haptic support with passive IPT did not result in less variable body sway compared to the active mode in the two most challenging conditions. As the variability of the interaction force was comparable, we can ascertain that the average interaction forces are not affected by an averaging artefact of extreme values.

Despite less physical support by the therapist, the balance reduction is still greater in the active mode, which corroborates our conclusion that participants received additional cues facilitating of body sway control.

4.3 Interpersonal coordination of postural sway

In the AP direction of sway spontaneous in-phase in both active and passive IPT was the prominent IPC pattern, which confirms observations in previous studies [14, 15]. IPC was strongest in the two most challenging sensory conditions and in the majority of sensory conditions passive IPT resulted in stronger IPC than active IPT, with the exception of the easiest condition. Possibly, active stabilization of the participant by the therapist was applied less frequently in the easiest sensory condition with passive IPT, therefore causing weaker IPC, compared with the active IPT mode, in which stronger interpersonal entrainment [33] could have driven IPC. Fingertip IPT has been reported to result in lower cross-correlation coefficients compared to shoulder IPT [17], which might indicate that the involvement of a greater number of movement degrees of freedom in both partners interpersonal haptic interactions amounts to generally weaker IPC.

The corresponding time lags of the maximum in-phase cross-correlation coefficients demonstrated an average lead of 164 ms by the participant's over the therapist's body sway fluctuations. This is surprising as previous studies reported zero lags [14, 17, 34]. In these studies, however, visual feedback of the partner's body sway was not available or restricted to peripheral vision, which could have allowed haptic feedback to dominate the IPC. In this current study, the therapist kept open eyes permanently to observe a participant's body sway. We speculate, that visual dominance caused the therapist to automatically adopt a reactive follow-

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er mode [35, 36]. We observed a similar leader-follower relationship in a forward reaching task, when visual feedback was available to the contact provider [37].

Conclusion

We described the effects of passive and active involvement for balance support in a therapeutic context. The passive mode demonstrated increased strength of the interpersonal coordination and the active mode decreased the postural sway of the participant to a greater extent. We suggest balance training could be more effective when both partners face each other. Being more involved in the interaction might enable the participant to spend more time in a challenging balance situation searching and practicing a successful postural strategy. This still needs to be further investigated.

Conflict of Interest

There are no conflicts of interest for any of the authors.

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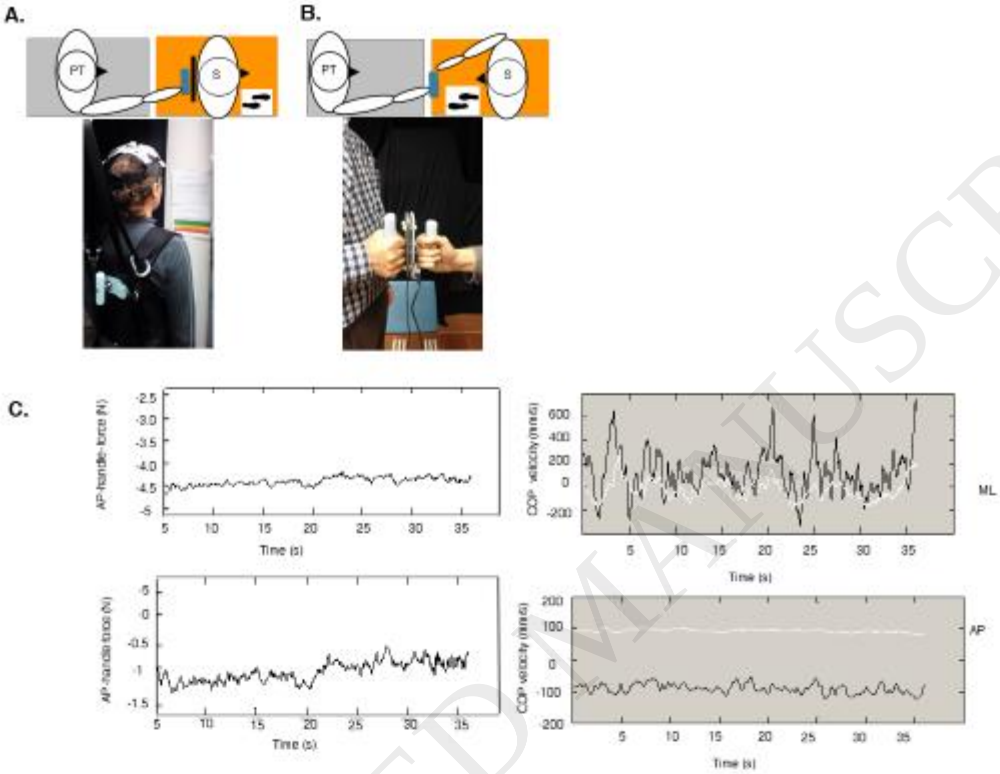
Fig. 1. (A & B) The stance configuration of the experimental setup at the beginning of a trial with the physical therapist on the grey force plate and the subject in semi-tandem on the orange force plate in the passive intermittent support mode (A) and in the active continuous support mode (B). The instrumented handle is represented by the blue rectangle in the schematic. Time series plots of the antero-posterior (AP) handle force (left) and AP and medio-lateral (ML) COP velocity of the physical therapist (light) and subject (dark) in active support mode during a foam surface , eyes closed and pitch movement trial (right) (C). The subject performs six balance exercises with increasing difficulty (D).



D.

	Surface	Vision	Head movement
1.	firm	eyes open	still
2.	firm	eyes closed	still
3.	firm	eyes closed	pitch
		Break	
4.	foam	eyes open	pitch
5.	foam	eyes closed	still
6.	foam	eyes closed	pitch

Figure1

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Fig. 2. The RMS COP velocity as a function of the exercise conditions and the support provision (passive/active) in AP (A) and in ML direction (B). Letters show the pairwise comparison between conditions; the same letters express conditions are not significantly different from each other. Bold dots indicate the significant support differences within each condition. Error bars indicate the standard error of the mean.

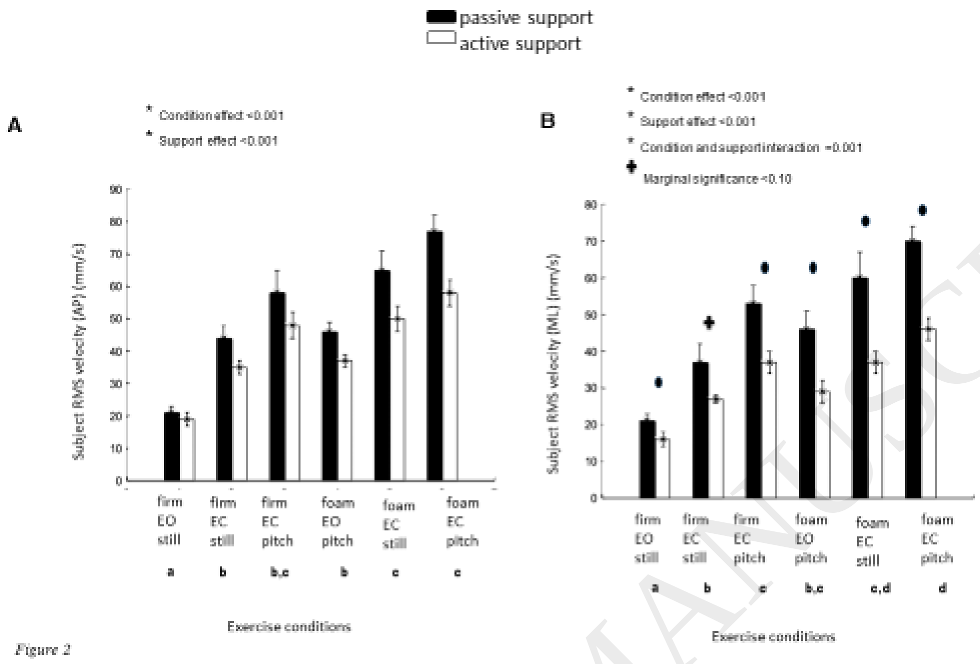

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Fig.3. The average (AV) of the handle force as a function of the exercise conditions and the support mode (passive/active) (A) as well as the standard deviation (SD) of the handle force as a function of the exercise conditions and the support mode (passive/active) (B). Letters show significant pairwise differences between conditions; same letters express that conditions are not significantly different from each other. Error bars indicate standard error of the mean.

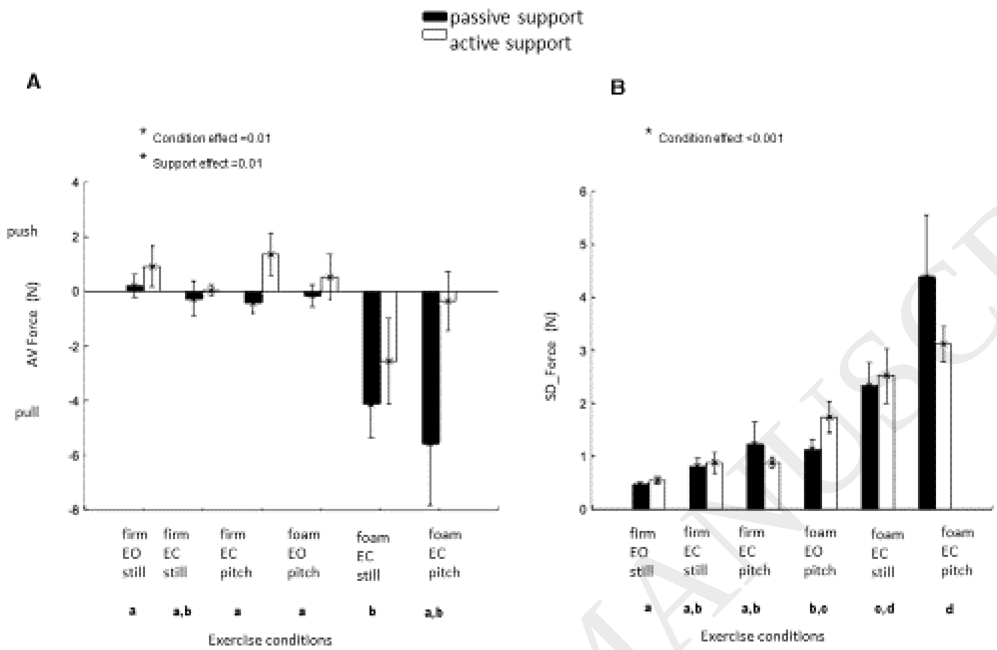

Figure 3

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Fig. 4. Upper panels show the average minimum cross-correlation coefficients of the CoP velocity as a function of the exercise conditions and the support mode (passive/active) in AP (A) and ML (B) direction. Lower panels show the average maximum cross-correlation coefficients of the CoP velocity as a function of the exercise conditions and the support mode (passive/active) in AP (C) and ML (D) direction. Statistical results refer to the Z-transformed cross-correlations. Minimum cross-correlations represent negative values and are shown rectified for better visual understanding. Error bars indicate the standard error of the mean. Letters show significant differences between conditions; same letters express conditions that are not significantly different from each other.

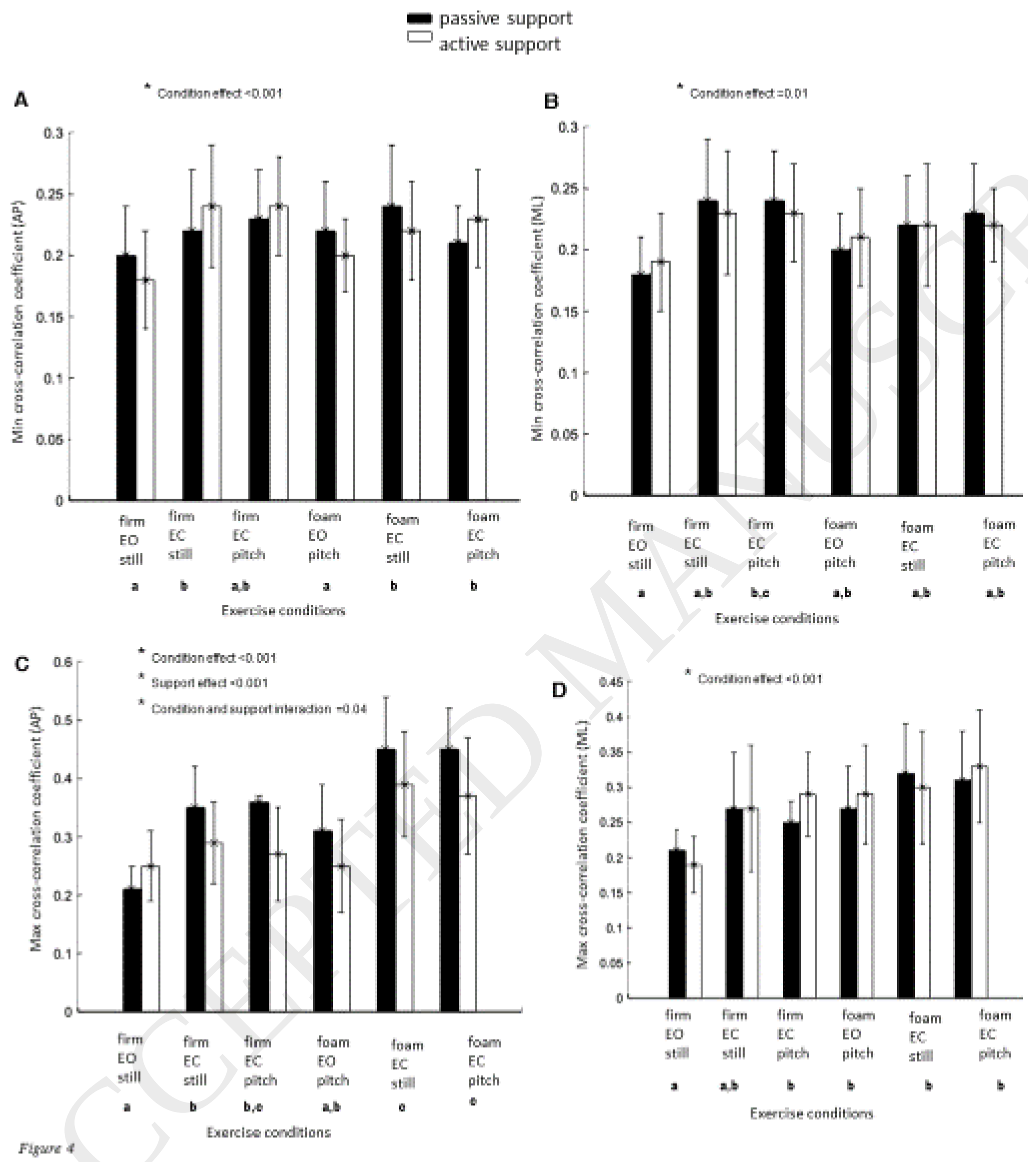

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Fig.5. Upper panels show the minimum average cross-correlation lags of the CoP velocity as a function of the presence of the exercise conditions and the support mode (passive/active) in AP (A) and ML (B) direction. Lower panels show the maximum average cross-correlation lags of the CoP velocity as a function of the presence of the exercise conditions and the support mode (passive/active) in AP (C) and ML (d) direction. Statistical results refer to the Z-transformed cross correlations. Error bars indicate the standard error of the mean.

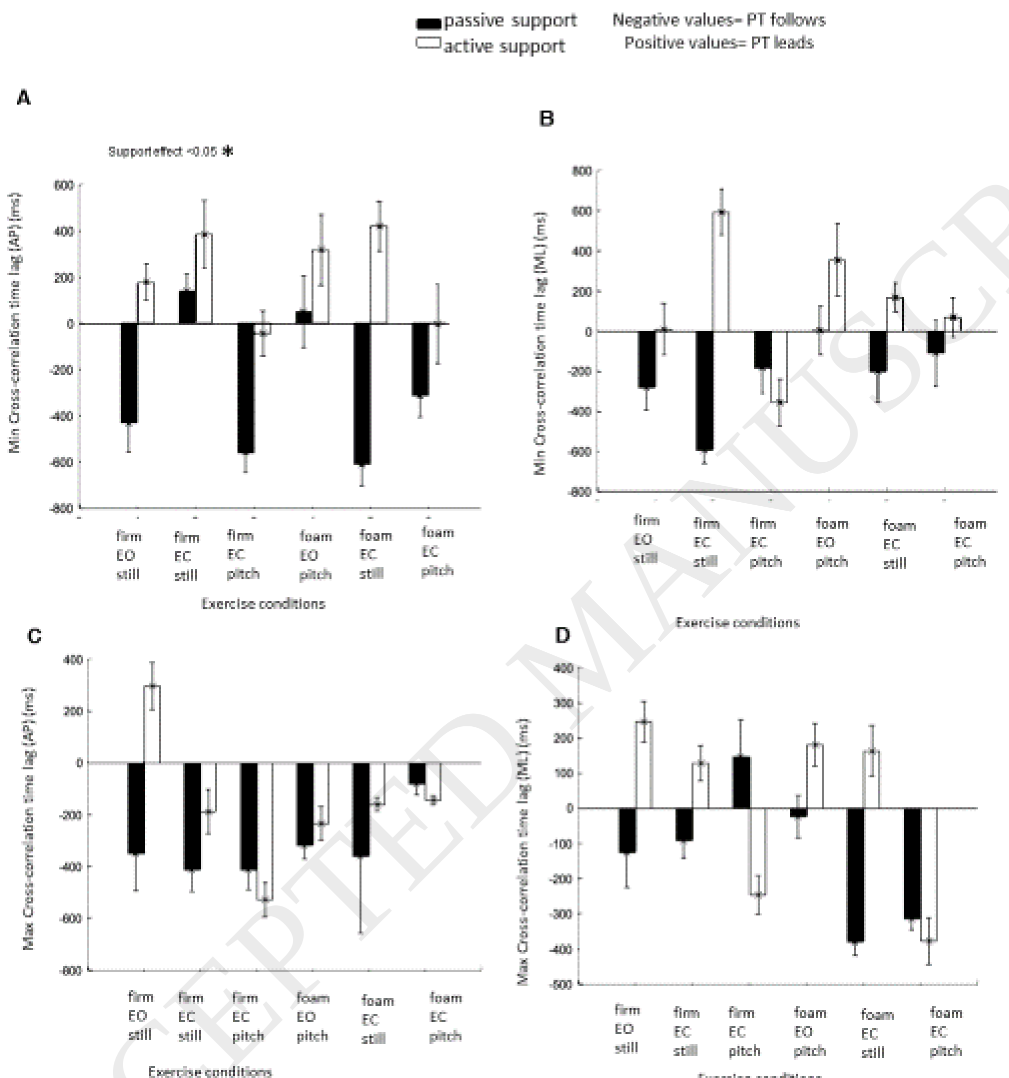

Figure 5

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