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Interpersonal interactions for haptic guidance during balance exercises

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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.

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,

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"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 disambiguation 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

2.1. 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 (16 years of experience) provided support.

2.2. 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.

2.3. 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).

2.4. Experimental design

Participants performed 2 sets of 6 randomized balance exercises during two different conditions: passive support (PS) and active support (AS) (Fig. 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].

2.5. 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 and B). A tri-axial load cell (DSA-03 A 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).

2.6. 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 6 cm, 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 s and participants wore a safety harness.

2.7. Data reduction and statistical analysis

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 = 10 Hz). 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.