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*Research*

# Light touch for balance: influence of a time-varying external driving signal

Alan M. Wing\*, Leif Johanssen and Satoshi Endo

*Sensory Motor Neuroscience Laboratory, School of Psychology, University of Birmingham  
Edgbaston, Birmingham B15 2TT, UK*

Sensory information about body sway is used to drive corrective muscle action to keep the body's centre of mass located over the base of support provided by the feet. Loss of vision, by closing the eyes, usually results in increased sway as indexed by fluctuations (i.e. standard deviation, s.d.) in the velocity of a marker at C7 on the neck, s.d. dC7. Variability in the rate of change of centre of pressure (s.d. dCoP), which indexes corrective muscle action, also increases during upright standing with eyes closed. Light touch contact by the tip of one finger with an environmental surface can reduce s.d. dC7 and s.d. dCoP as effectively as opening the eyes. We review studies of light touch and balance and then describe a novel paradigm for studying the nature of somatosensory information contributing to effects of light touch balance. We show that 'light tight touch' contact by the index finger held in the thimble of a haptic device results in increased anteroposterior (AP) sway with entraining by either simple or complex AP sinusoidal oscillations of the haptic device. Moreover, sway is also increased when the haptic device plays back the pre-recorded AP sway path of another person. Cross-correlations between hand and C7 motion reveal a 176 ms lead for the hand and we conclude that light tight touch affords an efficient route for somatosensory feedback support for balance. Furthermore, we suggest that the paradigm has potential to contribute to the understanding of interpersonal postural coordination with light touch in future research.

**Keywords:** sensory; motor; balance

## 1. INTRODUCTION

Upright bipedal stance is an inherently unstable posture. In the sagittal plane, centre of mass (CoM) of the body naturally lies in front of the ankle, so the tendency is for anterior (forward) sway. In the frontal plane, the bridge-like frame formed by the legs and pelvis with CoM in the middle appears more stable, however, the slightest sideways displacement in either direction results in sway in that direction. In either case, muscle action (for example, the calf muscles in the case of forward sway or left or right hip abductor in the case of lateral sway) is required to arrest and reverse the sway. Thus, standing is a matter of correcting the tendency to sway by activating appropriate muscles.

Changes in the level of muscle activation to maintain upright stance produce fluctuations in ground reaction forces and torques, and these may be used as an index of the degree of control required for balance. Typically, the measure employed is the variability of the rate of change of the centre of pressure (s.d. dCoP), where CoP reflects the point through which the net forces

and torques act in anteroposterior (AP) and mediolateral (ML) directions [1]. Normally, multiple sensory cues, including vision, vestibular sensation, proprioception (leg muscle) and tactile sensations (soles of the feet) are available to the nervous system for the detection of sway. Reduced or absence of input to any of these sensory channels, but especially vision, often results in increased sway excursions that require greater muscle activation to compensate, and this is indexed by greater s.d. dCoP. However, in the case of loss of vision, light touch contact with just the tip of one finger with a stable environmental surface restores sway (or s.d. dCoP) to the level associated with full vision. In this paper, we review a series of studies from the last 15 or so years identifying sensory factors contributing to light touch attenuation of sway. We then describe results from a new paradigm, involving movement of the contact point, which we use to explore the underlying mechanism of light touch contributions to balance.

### (a) *Review of studies of the effect of light touch on sway*

In the first of many publications relating to the light touch paradigm, Lackner and co-workers [2,3] asked participants to stand heel to toe (tandem Romberg stance) which increases ML sway with the right-hand lightly touching a waist-high force transducer on one

\* Author for correspondence ([a.m.wing@bham.ac.uk](mailto:a.m.wing@bham.ac.uk)).

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One contribution of 18 to a Theo Murphy Meeting Issue 'Active touch sensing'.

side. The transducer was connected to provide an audible warning if the vertical force exerted by the participants exceeded 1 N. In one condition the eyes were open and in the other they were closed. Side-to-side sway in the frontal plane was reduced with light touch to the same low level whether the eyes were open or closed. In an additional condition ('force touch'), participants were allowed to use as much normal force as they wished, typically three or four times more than in light touch. This condition yielded low levels of sway similar to light touch. However, the two conditions differed in the correlation between shear force at the finger and ML CoP fluctuations. In the light touch condition, the peak correlation coefficient was +0.6 and the lag a little over 350 ms, with shear force at the hand leading CoP. In the force touch condition, the correlation was larger and the lag was smaller. The authors suggested that, in the light touch condition, the shear force at the finger drove postural corrections to maintain light touch with low normal force. The efficiency of this additional tactile feedback route, compared with normal proprioceptive and tactile channels available in the absence of vision, resulted in reduced sway. In contrast, it was suggested that the force touch condition reduced sway by physically stabilizing balance. However, the degree to which arm weight is more or less taken on the transducer does not itself determine whether the force condition results in mechanical stabilization. Instead, this depends on how stiff the kinematic chain is linking the arm to the body. We suggest that an alternative possible account of the differing lag between light and force touch is that the heavier contact provides clearer sensory information about sway allowing faster and more accurate compensatory balance adjustments.

The initial demonstrations by Lackner and colleagues of the benefits of light touch on balance involved the tandem Romberg standing posture, which is particularly unstable in the frontal plane. Subsequently, Clapp & Wing [4] obtained a similar effect of light touch on balance in the sagittal plane with normal bipedal stance. They also observed a positive correlation between hand shear force and CoP with lag of 350 ms in support of a tactile feedback loop reducing sway. If the effect of light touch on balance derives from additional feedback, reducing feedback should remove the benefits. Reginella *et al.* [5] asked participants in normal stance to make light touch contact with a vertical surface using the index finger. While the authors found that light touch contact reduced sway when the vertical surface was fixed, they found AP sway with contact surface that moved simultaneously with own AP sway, was as large as, or larger than, the sway with no contact. This result suggests that reducing the information about sway relative to the contact point provided by the tactile signal impaired the ability to compensate for AP sway. An even more direct approach to manipulate tactile feedback was taken by Kouzaki & Masani [6], who observed that sway reduction with light touch disappeared when the hand was anaesthetized using a compression block on the upper arm.

Further evidence of the utilization of tactile feedback from light touch in standing balance was

demonstrated in a paradigm in which the contact surface was oscillated rather than being fixed. Jeka *et al.* [7] asked participants standing in tandem Romberg posture to place their finger lightly (vertical force less than 1 N) on a surface that oscillated sinusoidally with a peak velocity of  $65 \text{ mm s}^{-1}$  in the frontal plane at one of a set of frequencies ranging from 0.1 to 0.8 Hz (amplitude ranged from 18 to 2.25 mm). For oscillations up to 0.5 Hz, spectral analysis of the sway showed the presence of the contact surface oscillation frequency. This indicates that oscillation of the finger entrained the postural corrections in a 1:1 manner as would be expected if the tactile input were being used as a reference signal in maintaining posture. Given entrainment, it might be expected that this would have elevated sway variability relative to no contact, however, the study did not include a no-contact condition and so this possibility remains open.

Light touch reduction of sway does not necessarily need to involve the hand and arm but also occurs with 'passive' light touch in which an environmental referent, in the form of a fixed flexible contactor covered in soft textured material, touches the skin. Rogers *et al.* [8] asked participants to stand with flexible soft contact of this kind at the leg or shoulder. They found that sway was reduced in both cases, and was reduced more with the shoulder than with the leg contactor, presumably because a given degree of sway results in greater variation in force, or excursion of the contactor, at the higher point on the body. Furthermore, Rogers *et al.* [8] found that the sway reduced more when both contactors were applied together, indicating summation of information from the two sources. In analogous manner, greater reduction of sway with two compared to one contact point was also demonstrated by Dickstein [9] for 'active' light touch with two versus one index finger making light touch contact.

In the studies reviewed so far, the participants' task was to stand quietly. However, standing balance is often assessed by the response to dynamic perturbation involving movement of the support surface [10] or a push to the body [11]. It is therefore interesting to ask whether light touch benefits recovery from dynamic perturbations to balance. Johannsen *et al.* [12] examined the effect of passive light touch at the left shoulder on the response to forward sway owing to balance perturbations produced either by voluntary, self-initiated or involuntary, experimenter-imposed pull on the right-hand tending to cause forward sway. They found that balance was restored faster (earlier return of sway to pre-perturbation levels) with passive light touch, both after self-initiated and experimenter-imposed perturbation.

In light touch-assisted balance, the contact surface need not be rigidly fixed to reduce sway. Thus, Rogers *et al.* [8] reported that there was a light touch effect, even in the more variable sway condition (standing on foam) in which there was sliding of the shoulder level contactor. Moreover, Riley *et al.* [13] observed that sway was reduced by contact with a curtain, although this benefit of light touch contact was obtained only when the participant's attention was drawn to keeping the finger in contact without disturbing the curtain. This study illustrates that the forces

involved in the light touch effect can be very light. Indeed, in a study by Backlund Wasling *et al.* [14], reduction in sway was obtained with an air jet directed at the pad of the index finger providing the light touch contact. This demonstrates that variation in the spatial location of contact without any shear force is sufficient to achieve the light touch effect. However, in cases of low or zero force it may be that the attention demands of using the light touch information are increased. Thus, Vuillerme *et al.* [15] showed that sway reduction using light touch with a curtain leaves less attentional capacity for a secondary auditory detection task (using reaction time as a measure) compared to standing with full vision, which would, presumably, have yielded comparable levels of sway. Light touch contact with a rigid surface might then be expected to be less attention-demanding than curtain contact, although, to our knowledge, this has not yet been tested. Moreover, other finger touch conditions that result in enhanced reduction in sway, such as force contact (cf. [3]) or holding the finger clipped to the contact surface [16] might also reduce the attention demands compared with light touch contact.

Recently, we have shown [17] that, even a moving contact surface, such as that provided by touching another person, who is also swaying of course, enables reduced sway with light touch contact. Participants stood side by side on two separate force plates with elbows flexed and the right or left hand held out forward of the trunk. They were instructed to stand still and, in different conditions, make light contact with each other's index fingers, with a fixed surface, or no contact. The interpersonal light touch (IPLT) contact condition resulted in a reliable reduction in sway compared with no contact, albeit the effect was smaller than that achieved with contact with the fixed support. One interpretation of this result is that, in IPLT, information about own sway from light touch contact is degraded by the contact surface movements owing to the sway of the other person. However, another aspect of the task, the need to follow the sway of the other person to maintain contact, may also be a factor in sway being greater than that with a fixed point of contact. This view is consistent with the finding that the ground reaction forces of the two participants were reliably correlated in the IPLT contact condition. Recently, we obtained further support for the role of contact point movement in a study [18] which included a shoulder to shoulder IPLT condition. We found that under conditions in which one person was in stable bipedal stance and the other was in unstable tandem stance, IPLT contact increased the sway of the person in stable stance and decreased that of the other person in unstable stance.

### (b) Outstanding issues

In summary, the various studies of light touch suggest that force and position information associated with light touch contact provide cues to balance that can be combined with other cues, such as visual or vestibular information, to determine the current postural state and take action to move towards a desired, more stable, state. The touch information appears to

be used to control body sway more efficiently than if there were no contact. The efficiency may be owing to more accurate, or earlier, motor commands owing to improved sensory information indicative of own body sway. Factors affecting that efficiency include contact attributes (e.g. spatial location, active versus passive, force level) and also attention. Sway can be entrained by oscillating light touch contact, and, if there is IPLT contact with another person, whether sway increases or decreases appears to depend on the relative stability of each individual.

Sway reduction with IPLT is a very interesting finding as this form of contact between people is widely observed, for example, in holding hands. Of course, the form of finger contact commonly involves a full grasp rather than the single finger contact of the laboratory paradigm. However, in holding hands, the grip and both partners' arms generally appear relaxed, so it is perhaps not unreasonable to suppose that there is normally very little net force between partners and that this joint posture would probably meet the criterion of 'light touch'. Given that IPLT in standing is so common, it is surprising that relatively little is known about the parameters of sensing and control in IPLT and the contribution to balance brought by this mode of light touch contact.

To investigate processing of finger light touch for arm movement and balance, we now describe an experiment in which participants, with eyes closed, made light touch contact between the index finger tip and a haptic device. Unlike most of the studies described above, in which the contact point could move, we followed the study of Krishnamoorthy *et al.* [16] in using a form of contact in which the finger was restrained or clipped to a contact surface. Their apparatus exerted a (normal) pinch force on the finger of 14 N, but with less than 1 N (tangential) pull force by the clip on the finger. As noted earlier, with this form of tight and light contact, Krishnamoorthy *et al.* [16] found that there was a greater reduction in sway than the normal light touch contact. We chose this 'light *tight* touch' form of contact in the hope of improving the efficiency of the linkage between somatosensory input and balance output. We therefore predicted positive cross-correlations between hand and ground reaction force with hand lead time shorter than the previously described 350 ms [3,4]. We measured postural sway (rate of displacement of a C7 marker, dC7) and balance adjustments (rate of change of centre of pressure, dCoP) as a function of a range of actuator movement conditions comprising: (i) No-contact control condition, in which the arm was held with the haptic device thimble on the finger but not connected to the drive mechanism, hence there was no force feedback. (ii) The Earth-fixed reference condition, replicating Krishnamoorthy *et al.* [16], which we expected would result in significantly reduced sway. (iii–v) Sinusoidal trajectories of the haptic device with frequencies of 0.3, 0.5 and superimposed 0.3 + 0.5 Hz, to replicate and extend the study of Jeka *et al.* [7]; we expected that entrainment in conditions (iii–v) would result in greater sway than in (i,ii). (vi) Biological movement contact condition with haptic trajectory selected from a set of trajectories

sampled from other individuals in the no-contact control condition. We considered condition (vi) as capturing one element of IPLT, extending the work of Johannsen *et al.* [17], and predicted that sway would be greater than the fixed condition (ii) but less than in the no-contact condition (i) and the moving support conditions (iii–v) because the non-periodic reference signal would not entrain sway.

## 2. METHODS

### (a) *Participants and procedure*

Nine participants (mean = 26.3 years, s.d. = 4.6 years; five females and four males, all right-handed for writing) were tested while standing in stockinged feet on a force plate (Bertec 4060H, OH, USA) in normal bipedal stance with 5 cm inter-heel gap, eyes closed and head facing forward. The force platform measured the six components of the ground reaction forces and moments to determine the AP and ML components of the centre of pressure (CoP). Participants were instructed to stand as still as possible in a relaxed manner without speaking and eyes closed. The right arm was extended with the elbow in contact with the torso at waist level while the left arm was brought across the stomach so that the other hand made contact with the crook of the extended arm. Written informed consent was obtained from all participants and the study was approved by the University of Birmingham Ethics Committee.

Body sway was recorded in six experimental conditions. In each condition, the participant's dominant index finger was kept in the thimble of a haptic device (PHANToM 1.5 Sensable Technologies, MA, USA). In all except the first condition, which served as a no-contact control condition, the thimble was engaged with the haptic device which was located 40 cm in front of the participant. The hand was held in pronation and the index finger was extended but relaxed so that passive movements of the finger joints by the haptic device were still possible. A virtual plane was implemented at the hip level beyond which the thimble could not be lowered without deliberate effort against a resisting spring force updated at 200 Hz. Participants were instructed to maintain constant 'light touch' on this plane. The thimble was free to move in the ML direction as well as upwards away from this planar barrier. The haptic device, however, controlled the thimble's position in the AP direction in an open-loop mode according to a pre-specified trajectory. The haptic device's force output was always limited to a maximum of 1 N and therefore the thimble could deviate from this trajectory if participants generated a force larger than 1 N.

In each of the five experimental conditions, the haptic device produced one of a number of different categories of thimble trajectory. In the remaining condition, the index finger was also kept in the thimble (mass 19 g) detached from the haptic device. This condition (i) served as a 'no-contact' (no force feedback) control condition with an equivalent 'finger in thimble' sensation. The five conditions with haptic stimulation were: (ii) thimble held by the haptic device at a constant position ('stiff' haptic device) with spring-like force

feedback directed towards the specific location on every axis (spring stiffness  $0.5 \text{ N mm}^{-1}$ ); (iii) sinusoidal 0.3 Hz oscillation; (iv) sinusoidal 0.5 Hz oscillation; (v) superimposed 0.3 and 0.5 Hz oscillations (SP); (vi) biological movement with playback of thimble movements during the no-contact control condition of one randomly chosen trial from each of five other individuals who were not taking part in the experiment (BL). The amplitude of the haptic device trajectory in each of the conditions (ii–vi) was scaled so that the standard deviation of the thimble position matched that in the no-contact condition (i). The average peak-to-peak amplitude across the participants was  $8.2 \pm 2.8 \text{ mm}$ . Each experimental condition was tested five times for a total of 30 trials. The no-contact condition was tested first in a block of five trials. The order of the remaining 25 trials was fully randomized. The duration of a single trial was 63 s, however, the first 2 s and the last second were removed so that only 60 s were analysed in each trial. Figure 1 shows the set-up and an illustrative trace for the stimulus input and the sway for each experimental condition.

### (b) *Data reduction and statistical analysis*

Data from the force platform, the haptic device and body movements at C7 captured by optical motion tracking (Qualisys Oqus, Sweden) were sampled at 200 Hz. Force platform recordings were processed to determine AP and ML components of CoP fluctuations. All data time series were smoothed using a 100 ms moving average window and differentiated to yield rate of change measures of sway (dC7, dCoP) and thimble velocity. Within-trial estimates of sway (s.d. dC7, s.d. dCoP) were subjected to ANOVA with experimental condition as within-subject factor. Significance levels were set at  $p = 0.05$  after Greenhouse–Geisser correction. The coupling between thimble movements and sway in the five conditions (ii–vi) involving the haptic device was analysed by calculating cross-correlation functions in the AP and ML directions. Cross-correlation functions were computed for time lags ranging from +3600 ms (haptic device leads) to –3600 ms (sway leads). The largest absolute cross-correlation coefficient and corresponding time lag were extracted. The cross-correlation coefficients were Fisher–Z-transformed [19] and also subjected to ANOVA with experimental conditions as within-subject factor.

In order to quantify entrainment of body sway in the oscillating haptic stimulus conditions, spectral analysis was performed on the thimble of the haptic device ('driving signal') and also on the C7 and the CoP position time series. For each variable, the fast Fourier transform (FFT) was calculated with a window length (8192 data points) of the nearest power of 2 smaller than the number of total data points per trial, which resulted in a frequency spectrum with a step size of 0.0244 Hz per bin. Frequency bins from 0 to 0.1 Hz were excluded from the subsequent frequency peak extraction algorithm to avoid the inclusion of slow drift effects commonly observed in quiet normal bipedal standing. Three frequency ranges ('harmonics') based on the FFT of the

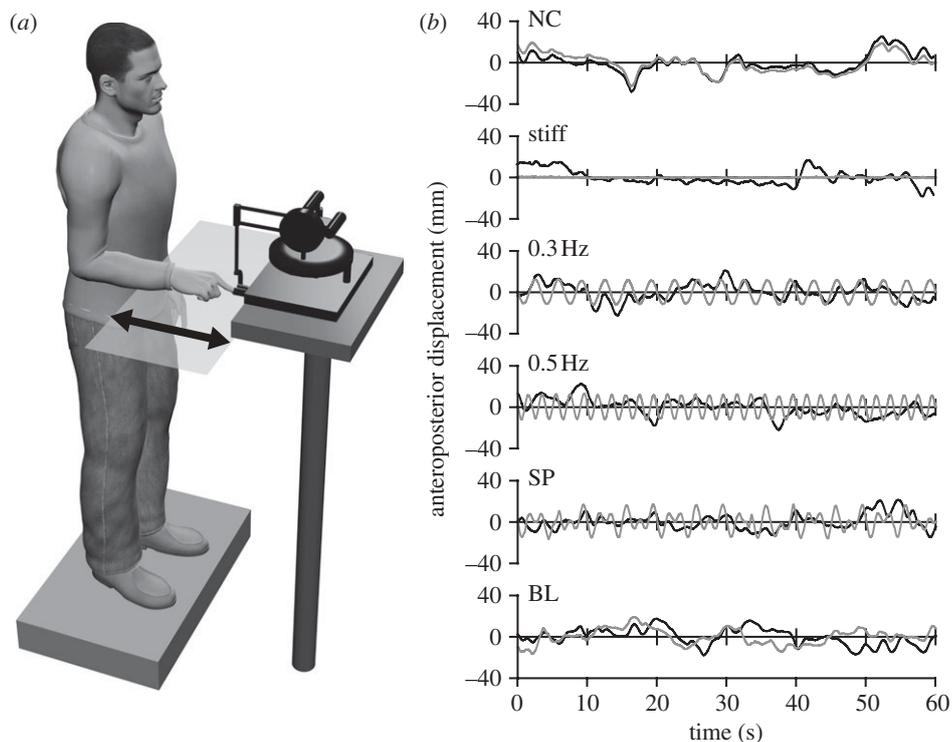


Figure 1. (a) The experimental set-up consisting of a participant in normal bipedal stance with the index finger of the right-hand inserted in the thimble of a PHANTOM 1.5 haptic device. The haptic device imposed six types of movements onto the finger along the anteroposterior axis (shown by the double-headed arrow). (b) Illustrative trajectories and sway (C7) fluctuations for a single trial in each experimental condition for a single participant. The light grey line shows the position change of the finger imposed by the haptic device. The black line represents C7 fluctuations during the haptic stimulation. In the stiff condition, a small amount of movement in the thimble was observed owing to the spring constant ( $=1$  N) used to prevent thimble displacement. NC, no contact; SP, superimposed periodic oscillation ( $0.3 + 0.5$  Hz); BL, playback of spontaneous biological thimble movements previously recorded from an individual not exposed to any resisting forces by the haptic device.

driving signal from the lowest frequency condition ( $0.3$  Hz) were defined for extraction of the respective peak frequency within each range. First, the frequency bin with the peak magnitude for the driving signal was found. Second, the range of the first harmonic was set up by adding and subtracting half the peak bin position. Then, the range for the second harmonic was defined by adding the width of the first harmonic range to the nearest frequency bin greater than the upper bound of the first harmonic and the same was done for the third harmonic range based on the upper bound of the second harmonic range. For every trial, the local peak frequency bin for the driving stimulus was located and the FFT magnitude over this bin as well as the phase were recorded within each of the three frequency ranges. For C7 and CoP, the FFT magnitude and phase were extracted from the same peak frequency bins identified for the driving stimulus. Finally, the frequency bin with the greatest absolute FFT magnitude was identified as the 'primary frequency' and the second largest was identified as the 'secondary frequency'. All data analysis was performed in MATLAB 7.5 (MathWorks, Natick, MA, USA) and SPSS 16 (IBM Corporation, Somers, NY, USA).

### 3. RESULTS

#### (a) Thimble force and movements

We first characterize the thimble force and movement parameters, as they can be affected by both the original

driving stimulus provided by the haptic device but also by the stiffness parameters of the participants' contacting finger. No difference was found between the experimental conditions with respect to the average force exerted by the haptic device, in contrast to variability of force where differences were apparent ( $F_{4,32} = 8.77$ ,  $p = 0.003$ ,  $\eta^2 = 0.52$ ). The s.d. of force was lowest for both the stiff and 'biological' conditions (mean,  $M = 0.25$  N, s.d. =  $0.13$ ) with a significant increase for the remaining conditions comprising the simple and complex oscillations ( $M = 0.43$  N, s.d. =  $0.14$ ;  $F_{1,8} = 32.83$ ,  $p < 0.001$ ,  $\eta^2 = 0.80$ ). Also, peak thimble velocity tended to differ between the experimental conditions ( $F_{4,32} = 3.24$ ,  $p = 0.09$ ,  $\eta^2 = 0.29$ ). Peak thimble velocity was lowest in the stiff condition which was induced by the participants' own finger movement exerting a force that temporarily exceeded a maximum  $1$  N of the haptic device ( $M = 10.5$  mm s<sup>-1</sup>, s.d. =  $4.7$ ), followed by the biological condition ( $M = 22.0$  mm s<sup>-1</sup>, s.d. =  $9.4$ ), the  $0.5$  Hz condition ( $M = 38.7$  mm s<sup>-1</sup>, s.d. =  $24.7$ ), the 'superimposed' condition ( $M = 41.3$  mm s<sup>-1</sup>, s.d. =  $22.7$ ) and the  $0.3$  Hz condition ( $M = 42.7$  mm s<sup>-1</sup>, s.d. =  $41.3$ ). Finally, the experimental conditions were different with respect to the s.d. of thimble velocity ( $F_{4,32} = 10.0$ ,  $p = 0.009$ ,  $\eta^2 = 0.56$ ). Thimble velocity was least variable in the stiff condition ( $M = 2.5$  mm s<sup>-1</sup>, s.d. =  $1.13$ ), followed by the biological condition ( $M = 5.9$  mm s<sup>-1</sup>, s.d. =  $2.6$ ), the  $0.3$  Hz condition ( $M = 12.2$  mm s<sup>-1</sup>, s.d. =  $4.8$ ) and the superimposed

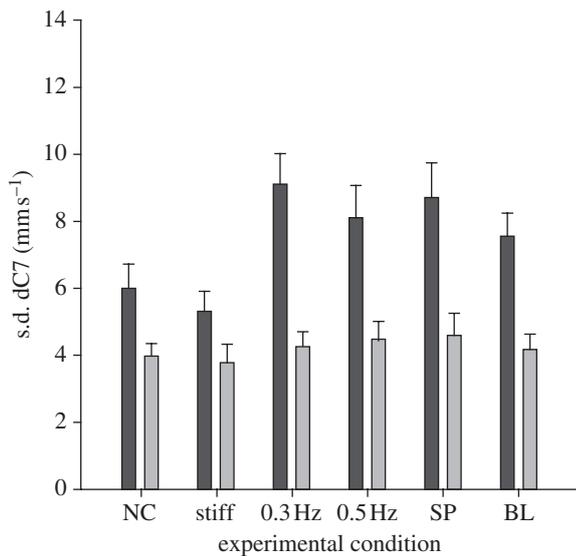


Figure 2. Sway (s.d. of C7 rate of change, s.d. dC7) shown as a function of experimental condition and sway direction. Error bars indicate the standard error of the mean. Black bars, anteroposterior; grey bars, mediolateral.

condition ( $M = 14.7 \text{ mm s}^{-1}$ , s.d. = 10.3). Variability of thimble velocity was greatest in the 0.5 Hz condition ( $M = 15.6 \text{ mm s}^{-1}$ , s.d. = 11.8).

### (b) Body sway

Sway analysis focused on dC7 variability. Sway in terms of s.d. dCoP is relegated to the electronic supplementary material as the results were broadly similar to s.d. dC7. Figure 2 shows dC7 sway for each experimental condition in both directions of sway. Sway was greater in the AP direction ( $F_{1,8} = 33.28$ ,  $p < 0.001$ ,  $\eta^2 = 0.81$ ). In the AP direction, participants' sway showed the greatest variability in the 0.3 Hz condition, where s.d. dC7 increased by 65 per cent compared with the no-contact condition. However, in the ML direction, the sway increased by only 14 per cent. The least amount of sway occurred in the stiff condition, when the haptic device was set to keep the thimble at a specific target position. The relative sway reduction in the stiff condition was 9 per cent in the AP and no change (0.4% increase) in the ML direction compared with the no-contact condition. There was a main effect of experimental condition ( $F_{5,40} = 5.79$ ,  $p = 0.008$ ,  $\eta^2 = 0.42$ ) and a significant interaction between experimental condition and sway direction ( $F_{5,40} = 6.31$ ,  $p = 0.004$ ,  $\eta^2 = 0.44$ ). In the AP direction, *post hoc* comparisons against the stiff condition revealed significantly greater sway for all conditions (all  $F_{1,8} > 5.59$ , all  $p < 0.05$ , all  $\eta^2 > 0.41$ ). Furthermore, *post hoc* comparisons against the no-contact condition resulted in a significant increase in sway for the 0.3, 0.5 Hz and superimposed conditions (all  $F_{1,8} > 6.65$ , all  $p < 0.03$ , all  $\eta^2 > 0.45$ ) with a tendency for an increase in the biological condition as well ( $F_{1,8} = 3.47$ ,  $p = 0.10$ ,  $\eta^2 = 0.30$ ). In the ML direction, no changes in sway were found compared with either the stiff or the no-contact conditions.

### (c) Cross-correlations

Figure 3 shows illustrative cross-correlation functions between dC7 and thimble velocity, averaged across all five trials, as well as overall peak cross-correlation coefficients for each of the experimental conditions on AP and ML axes. Concerning dC7 on the AP axis, cross-correlation functions show a single positive peak with slightly positive phase lag, indicating that the haptic device led the sway of the participants during the stiff and the biological conditions. In the periodic oscillation conditions, a positive peak with a slightly positive phase lag can also be seen. The cross-correlation functions, however, show gradually damped oscillation as the lag departs further from the peak. Peak correlation coefficients between the thimble velocity and both sway measures were exclusively positive indicating an in-phase coupling between the two variables. On the ML axis, cross-correlation functions for dC7 showed a single peak at either short positive or negative phase lags, depending on the experimental conditions. Peak coefficients of the cross-correlation function between dC7 and thimble velocity were significantly lower for the ML direction ( $F_{1,8} = 51.29$ ,  $p < 0.001$ ,  $\eta^2 = 0.87$ ). Correlations with dC7 showed no differences between the experimental conditions, although the interaction between experimental conditions and sway direction tended towards significance ( $F_{4,32} = 2.87$ ,  $p = 0.08$ ,  $\eta^2 = 0.26$ ). For the AP axis, *post hoc* comparisons between the stiff and the remaining experimental conditions revealed a marginally lower correlation coefficient for the 0.5 Hz condition ( $F_{1,8} = 4.27$ , both  $p = 0.07$ , all  $\eta^2 = 0.35$ ). Peak correlation phase lags were statistically different between the two sway directions for dC7 ( $F_{1,8} = 15.30$ ,  $p = 0.004$ ,  $\eta^2 = 0.66$ ). In the AP direction, dC7 significantly lagged behind thimble velocity by 176 ms (s.d. = 92;  $t_8 = 5.73$ ,  $p < 0.001$ ). On the ML axis, average time lags were not reliably different from zero ( $M = 62 \text{ ms}$ , s.d. = 160).

### (d) Periodic haptic entrainment

Figure 4 shows the FFT magnitudes for both the position of the thimble of the haptic device and C7 as a function of each of the three periodic driving stimuli as well as the primary and secondary frequency for the 0.3, 0.5 Hz and superimposed conditions. For the 0.3 Hz condition, the thimble consistently showed a primary frequency at 0.29 Hz and for the 0.5 Hz condition, the primary frequency lay at 0.40 Hz. The superimposed condition showed primary and secondary frequency components at 0.28 and 0.45 Hz, respectively. FFT magnitudes at the driving frequency were significantly lower for C7 than for the driving stimulus ( $F_{1,8} = 6.58$ ,  $p = 0.03$ ,  $\eta^2 = 0.45$ ).

## 4. DISCUSSION

Light touch contact in which the hand is held with controlled force against an environmental surface stabilizes balance [3]. To investigate processing of finger light touch for the control of arm movement and balance, we have described a novel experimental paradigm in which participants stood with eyes

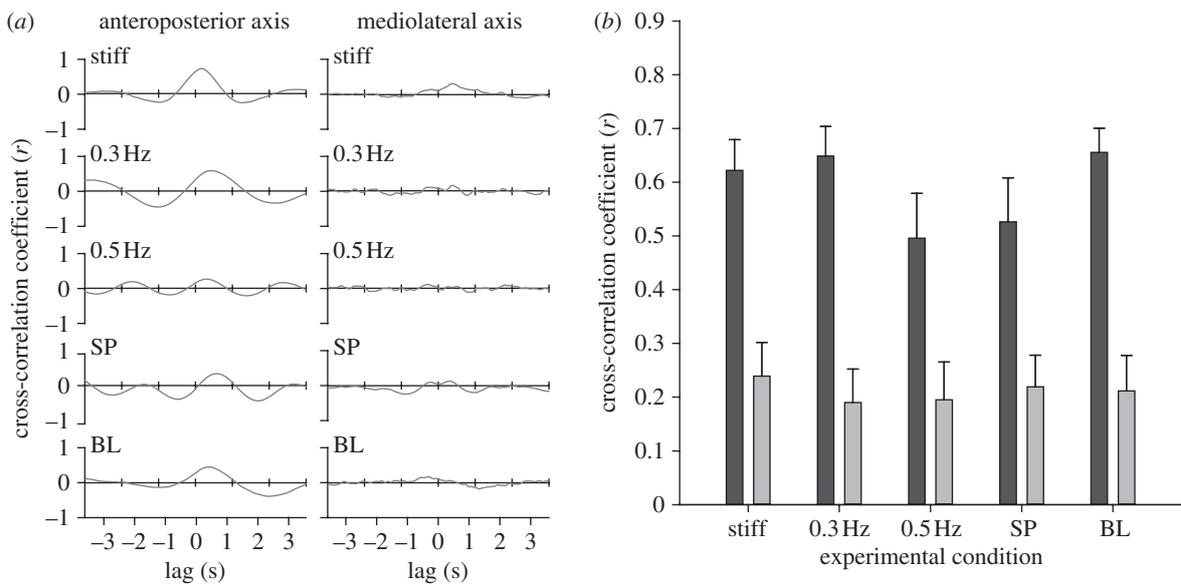


Figure 3. (a) Illustrative cross-correlation functions between dC7 and thimble velocity, averaged across the five trials, for each experimental condition and sway direction. Negative phase lags indicate a lead of dC7, while positive phase lags indicate a lead of the thimble velocity. (b) Peak correlation coefficients as a function of experimental condition and sway direction (black bars, anteroposterior; grey bars, mediolateral axis).

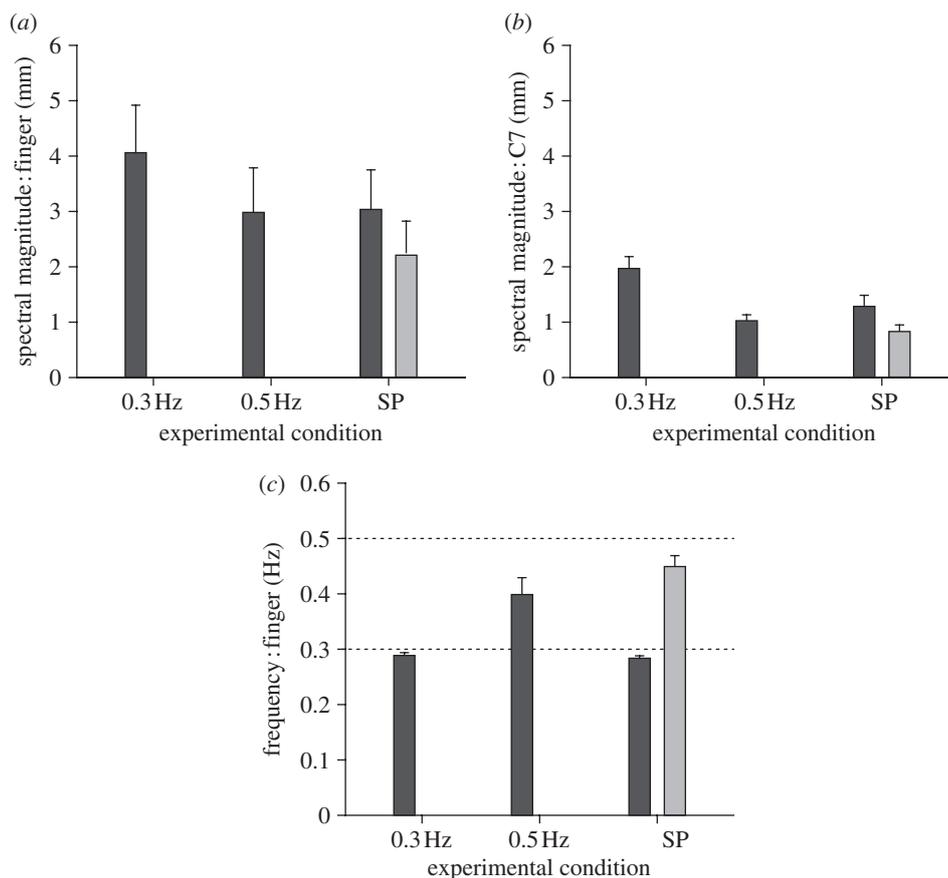


Figure 4. Peak spectral magnitudes over the respective frequency bin for the displacement of the thimble of the haptic device (a) and C7 fluctuations (b) for the three periodic oscillating conditions. (c) Frequency bin of the primary (black bars) and secondary (grey bars) frequencies embedded in the movements of the driving stimulus. The dotted line indicates the idealized target frequencies from which deviations were possible depending on the force exerted by the participant on the thimble.

closed and with the index finger tip placed in the thimble of a haptic device creating light tight contact. We examined two sets of conditions, one set providing balance cues expected to entrain balance and increase

sway, the other set providing balance cues expected to reduce sway.

In the conditions expected to entrain and increase sway compared with the no-contact condition (finger

still in the thimble, but the thimble disconnected from the haptic device), lower (0.3 Hz) and higher (0.5 Hz) frequency sinusoidal movement of the haptic device resulted in an increase in sway (s.d. dC7). However, in both sinusoidal movement conditions, the analysis comparing frequency components of the thimble of the haptic device and of the sway, showed evidence of entrainment, with similar proportions of power evident in C7 relative to the thimble for both conditions. In the condition with the 0.3 Hz frequency present in the motion path of the thimble, the greatest increase in sway was seen compared with the no-contact condition. Also in the superimposed condition where both 0.3 and 0.5 Hz were present, there was evidence of entrainment by the component frequencies, with similar proportion of power in the combined condition as for the low- and high-frequency conditions. Thus, our results with light tight touch contact replicate and extend the study of Jeka *et al.* [7].

In the conditions with light tight contact expected to reduce sway compared with 'no contact', we observed a reduction in sway in the stiff condition (finger held fixed on all three axes) only. In the biological condition (haptic device replayed the previously recorded sway path of another participant), sway was increased compared with the stiff condition but not as much as in the 0.3, 0.5 Hz or superimposed conditions. This finding of an increased level of sway is in contrast to the result we previously obtained for actual IPLT [17]. While the previous study used a real person as a partner, the haptic device in this experiment produced a pre-recorded 'other' trajectory in an open-loop manner. That is, in the present study, the participant had sole responsibility (as opposed to sharing responsibility in the previous study) of following the movement in order to keep the contact force light. This might have somewhat elevated the sway level in the present study. It also shows, however, that the postural coordination between two partners with finger light touch contact may be subject to more complex internal dynamics than just the passive entrainment by the other person's sway.

In addition to examining sway across all conditions, we also evaluated cross-correlation between the velocity of the haptic device thimble and dC7. In the AP direction, we found strong correlations in the same direction with the thimble velocity leading by 176 ms. This value is considerably less than the previously reported leads of 350 ms. We attribute this to the tight contact between the finger and the thimble resulting in a clear sensation, which Krishnamoorthy *et al.* [16] previously noted appeared more efficient than light contact in reducing sway, although they did not report cross-correlations and lags. The latency value obtained is longer than the 120 ms typically associated with postural reflexes [20], which are usually identified with supraspinal pathways. However, they are in the range of haptic reaction times of 140–190 ms reported in the context of a manipulator control task [21]. Compared with the manipulator task, it would be reasonable to suppose that some additional time would be required for cortically mediated transformations mapping hand coordinates to whole body posture to render effective the postural adjustments evidenced in the dC7 fluctuations.

In all experimental conditions, except for the stiff condition, the driving stimulus provided by the haptic device was along the AP axis. In the stiff condition, a 1 N force was applied on all axes in order to resist any movements of the thimble away from the set position. In general, we would have expected greater cross-correlation coefficients than the ones presently seen in all conditions if ML thimble velocity were determined entirely by ML dC7 sway as suggested by the near-zero phase lags. Thus, it may be possible that the oscillating stimulus on the AP axis also affected ML sway.

The present study demonstrates an advanced methodology for probing the interaction between time-varying tactile stimulation at the index finger and continuous sway adjustments during upright standing. The use of such a programmable haptic device allows, not only the application of complex periodic oscillations in open-loop mode, but, in future could also be used for closed-loop interactions between the haptic device and a participant. For example, the haptic device could be programmed to respond adaptively to the position information from the participant in order to simulate the feedback control understood to be used by the participant. Such a paradigm might be extended further to a two-person task by functionally linking two haptic devices, each one serving as light touch contact for one person, passing information about the sway of the other person. The experimenter could then manipulate the virtual linkage between the two participants to better explore IPLT and its effects on balance.

In conclusion, we have reviewed an active field of research in which light touch contact contributes to the maintenance of stable balance. We have presented a new paradigm which allows the nature of the touch stimulus to be manipulated to increase or decrease sway compared with no-contact conditions. Using the paradigm, we have demonstrated reduced sway with fixed light tight contact compared to no contact, and increased sway when the light tight contact was subject to simple or superimposed sinusoidal oscillations or when it reproduced the sway path of another person. Cross-correlation between finger motion and sway in the AP direction was higher and showed shorter lags than in previous studies in which the contact was light (and potentially free to move) rather than held tightly (but not allowing AP forces above 1 N) as in the present study. We speculate the difference may reflect more efficient (and less attention demanding) processing in using the somatosensory input at the hand. In current studies, we are using this approach to explore quantitative models for the exchange of information between two people who allow joint improvement of their balance in IPLT.

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