Standing stability in the frontal plane determined by lateral forces applied to the hip

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Summary
A method for disturbing standing balance using controlled horizontal forces at the hips is described. Its use is illustrated by two experiments evaluating the effect on hip position of sideways force applied for a fixed period of 5 s. In the first experiment increasing sway in the frontal plane was observed with increasing force (12.5, 25 and 37.5 N). In the second experiment it was observed that sway decreased with stance width. In both experiments there was greater sway at the onset of force than at its termination. The results suggest that the method may offer a simple and reliable method of evaluating the efficacy of neural mechanisms involved in the maintenance of standing balance.

Key words: Stance, stability, posture, frontal plane, balance, perturbation, sway

Gait & Posture 1995; Vol. **3**: 38–42, March

Introduction

A common approach to assessing standing balance involves determining the consequences of a specified perturbing force. In some tests the destabilizing forces arise from the subject's own action, as in raising the arm. Since the subject produces his or her own perturbation to balance, such a manoeuvre tests the subject's ability to anticipate and minimize the disturbance. For example, when the arm is raised rapidly, anticipatory postural adjustments are seen as compensation for dynamic loads arising from the arm movement[1-6]. There are also compensatory postural adjustments for the long-term change in centre of mass position associated with holding the arm raised[7]. These may be planned in advance or they might be organized in reaction to altered sensory cues.

Another approach to the assessment of standing balance involves the experimenter introducing the force that perturbs equilibrium. Generally such tests are concerned with reactive mechanisms of the central nervous system, and often the conditions are designed to preclude anticipation by subjects. For example, a paradigm developed by Nashner and in common use today uses transient horizontal movement of the support base[8]. The resulting sway triggers patterns of stabilizing muscle activity that are sensitive to equilibrium conditions[9].

Moving the support surface is an indirect method for destabilizing the upright body compared to the commonly employed clinical practice of observing the effect of a push to the upper body. A protocol to make subjective judgments based on the latter method more reliable was described a number of years ago[10]. A modified version of the test that affords quantitative objective assessment was subsequently advanced by Lee et al.[11]. In their method standing balance was assessed in terms of the maximum steady horizontal load applied at hip level that the subject could withstand without hip movement. However, the test only involved static loads. In the present paper we document a method in which we examine the subject's response to changes in horizontal load applied at the hips. Our equipment is based on a system that was developed to set up stabilizing forces to assist a stroke patient maintain upright stance[12]. Our technique involves measuring the displacement of the hips when moderate destabilizing forces are applied horizontally at the hips. We also assess the hip displacement when the force being applied is termi-

Received: 29 September 1993
Accepted: 23 June 1994
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0966-6362/95/$9.50

nated (so that the subject no longer needs to resist the force).

Moving platform studies of balance control are usually constrained by the speed and travel distance of the platform. A trial usually comprises a short period of acceleration followed immediately by deceleration. The inertial forces acting to destabilize the subject therefore do not last for very long before they reverse direction. In contrast, the method we describe allows good temporal separation of the onset and termination of force. Thus we are able to compare the way subjects activate resistive forces for the onset of a destabilizing force and how they inhibit such forces when the destabilizing force is terminated.

We illustrate our approach with two experiments using healthy adults to explore the regulation of LR stability (hip displacement or sway in the frontal plane). In the first experiment we consider the effects of increasing the level of the applied force. We show increase in sway with increase in force. In the second experiment we evaluate the effects of stance width. We show that there is a reduction of sway with greater stance width. This work extends an earlier report in which the method was used with only one level of force and with one stance width to evaluate stroke patients' stability in the frontal plane[13].

Methods

Subjects

Twelve subjects, mean age 36.4 years (range 20–56 years), height 172.3 cm (range 158–187 cm) and weight 69.7 kg (range 52.5–91.5 kg), participated in the research. Ten took part in experiment 1. Eight of these subjects plus two others proceeded to experiment 2. None of the subjects reported any problems with balance.

Apparatus

The apparatus used to perturb hip position comprised two actuators (Type SMX01, Electric Actuator, Bradford, UK), aligned at 90° to each other to give left–right (LR) and anterior–posterior (AP) motion under force-servo control providing forces in the range -100 +100 N (see Figure 1). The actuators could be positioned vertically at the subject's waist height and were coupled to the subject at the level of the iliac crest using a moulded thermoplastic belt held closed with a Velcro strap. Each actuator consisted of a torque motor coupled through a lead screw to a force sensor (Type F421, Novatech, Hastings, UK). They allowed a 0.3 m range of movement with a max speed of 0.3 m s^{-1}; the step increases in force used in the experiments described below took approximately 200 ms. In the experiments described in this paper only the LR actuator was used to provide perturbing forces; the set force level for the AP actuator was zero and so it allowed free motion throughout.

An Apple Macintosh IIfx computer with a National

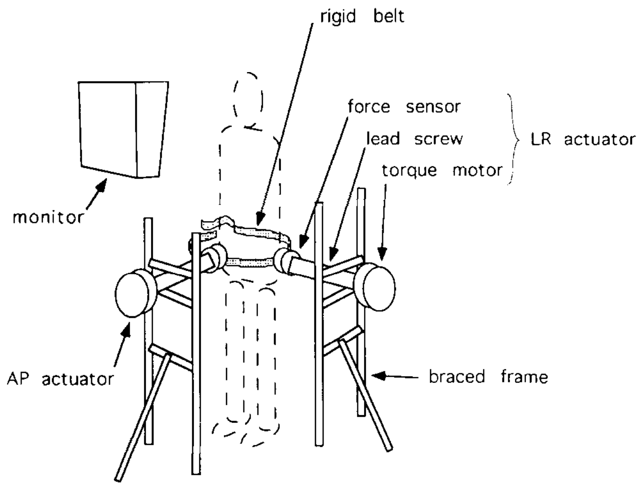

Figure 1. Apparatus to provide horizontal destabilizing forces at the pelvis. The monitor provides the subject with visual feedback about hip position.

Instruments analogue interface (NB-MIO-16) controlled the force levels of the actuators and sampled force and hip position at 50 Hz. A switch was placed convenient to subject and experimenter so that at any time computer control could be overridden and the actuators immobilized. The computer display was placed at the subject's eye level and provided a continuous indication of direction and time of the impending push (using a stopwatch-style display with sweep hand marking seconds). A horizontal slide marker gave feedback of left-right hip position.

Task

The subject was instructed to stand with hands to the side and with heels at a predetermined distance (this distance was monitored by the experimenter). Each trial started when the subject indicated that he or she was ready with hip position centred, as indicated on the computer display. Subjects were instructed to stand normally and resist the tendency of the hips to move sideways as the force was applied (push), 3 s into the trial, and then, 5 s later, removed (release). The trial terminated 5 s after this giving a total trial duration of 13 s.

Experimental design

After four familiarization trials, subjects were given six push-and-release trials, with the push directed alternately to the right and left. In Experiment 1 the force level was increased over successive blocks over the values 12.5, 25, 37.5 N. In this experiment, stance width (the distance between the heels measured in the middle over the point where the Achilles tendon inserts into the calcaneus) was 120 mm. In Experiment II the magnitude of the perturbing force was kept at 25 N and stance width was increased between blocks of trials from 50, 120, 190 to 260 mm.

Results

Figure 2 shows the typical form of hip displacement as a function of time. The change in force resulted in a rapid out-and-back sway movement of the hip; on push the sway was initially in the direction of the force, on release it was in the opposite direction. In the intervening period and at the end of the trial the hip reverted to a position close to that at the beginning of the trial. Visual observation by the experimenter (which was confirmed by viewing videorecordings made subsequently under similar conditions) indicated that hip (or more accurately, pelvis) motion was accompanied by motion of the knees in the frontal plane parallel to that of the hips but of smaller amplitude. Shoulder movement, which was also limited to the frontal plane, was generally equal to or less than the hip movement amplitude (i.e. there was a degree of side flexion of the upper body).

After conditioning the hip position data (a 2nd-order Butterworth low-pass filter with cut-off at 5 Hz was applied, and then every fourth data point used) a computer algorithm was used to determine three measures on each trial:

Peak displacements: the extreme values (measured relative to the initial position) in the direction of the force on push, and in the direction opposite to the force on release. Allowance for push direction was made by taking the absolute values of the peak displacements.

Return times: the time taken to return to the value of the initial position after the peak displacements on push and release. If the hip position failed to return to the initial position after push or release, a ceiling return time of 5 s was assigned.

Offsets: the average position over 1-s periods taken relative to the average lateral position of the hips over a 1-s period immediately prior to force onset (initial position). One offset was determined for the end of the push phase just before termination of force (offset on push), and the other for the end of the trial (offset on release). A sign convention was adopted that identified displacements towards the right as positive and towards the left as negative.

Experiment I: force level

Peak displacement. As push force increased each subject swayed further both on push and release. Figure 3 shows the peak displacement as a function of force for push and release. Three-way, repeated-measures analysis of variance (ANOVA) was carried out on the peak displacement with force (12.5, 25, 37.5 N), direction (left, right) and phase (push, release) as factors. This revealed the increase in peak displacement with force to be statistically reliable ($F(2,18) = 35.34$; $P < 0.01$). The difference between push and release was also reliable ($F(1,9) = 5.23$; $P < 0.05$). However, the apparent increase in difference between push and release with larger forces, i.e. the interaction between force and phase, was not statistically significant.

The functions relating peak displacement to force for the group data in Figure 3 appear straight. Non-significant contrasts between the 12.5 N condition with the 37.5 N condition and the 25 N condition confirmed that there was no statistically reliable departure from linearity for either push or release conditions. The linearity of the relation between peak displacement and force also held for each subject (as determined by testing for each subject the contrast between the 12.5 N condition with the 37.5 N condition and the 25 N condition). Correlations between peak displacement and force were computed for each subject and the averages for push and release were 0.89 and 0.93. The correlation across subjects of the slope of the displacement *versus* force functions was negatively correlated with body weight; −0.68 for push, −0.72 for release.

Return time. There was a statistically significant main effect of push *versus* release on return time ($F(1,9)$ =

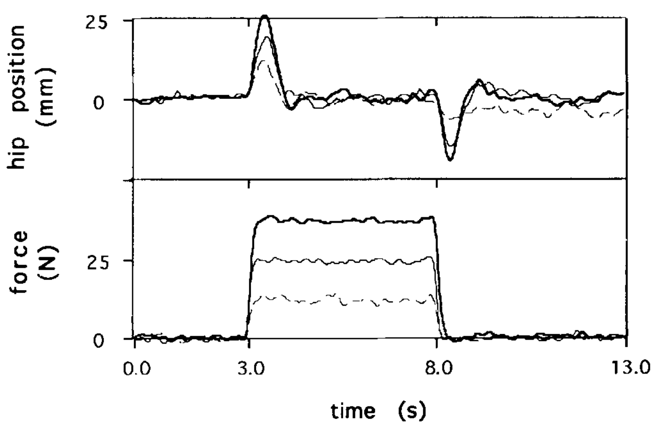

Figure 2. Single trial sample records from one subject showing hip displacement (above) as a function of time (sample interval 50 Hz) when a lateral force (below) is applied for 5 s starting 3 s into the trial. Sway increases across the three different levels of force shown below (12.5, 25 and 37.5 N, dashed, thin, thick lines).

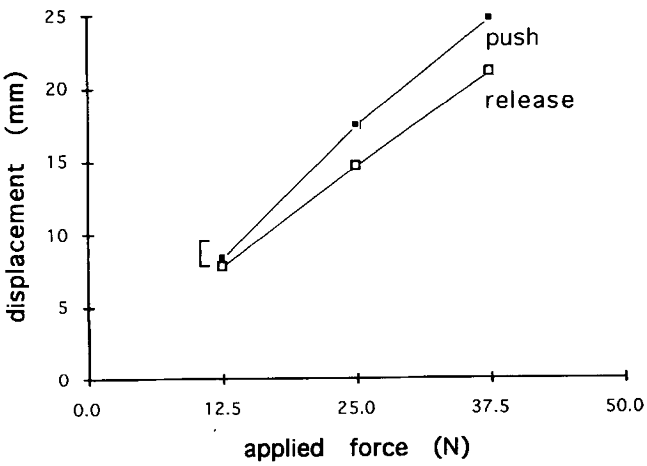

Figure 3. Experiment I. Increase in peak displacement with force level on push by and release from laterally directed force at the hip. Group mean data (with vertical bar indicating two standard errors) for 10 subjects.

36.34; $P < 0.01$) with the mean time for push (1588 ms) being considerably longer than that for release (919 ms). There was no reliable effect of force on return time.

Offsets. The offsets of push and release were close to zero. However, ANOVA revealed a statistically significant effect ($F(1,9) = 7.92$; $P < 0.05$) of phase (a 2.1-mm offset after push was reliably greater than a 1.5-mm offset on release). There was no reliable increase in offset with force.

Experiment II: stance width

Peak displacement. Subjects swayed less when pushed while standing with a wider base of support. Figure 4 shows peak displacement as a function of stance width for push and release. ANOVA showed the effect of stance width was significant ($F(3,27) = 7.95$; $P < 0.01$). There was also a reliable difference between push and release ($F(1,9) = 15.39$; $P < 0.01$). As in Experiment I, the peak displacement was smaller on release than on push. There was no significant interaction between stance width and phase.

Return time. Although there was a trend for a reduction in return time with increase in stance width the effect did not attain statistical significance. The difference between push (1539 ms) and release (1061 ms) was reliable ($F(1,9) = 12.25$; $P < 0.01$).

Offsets. Although offsets were small, there was a reliable main effect of stance width ($F(3,27) = 3.94$; $P < 0.05$) with the largest offset (2.5 mm) for the narrowest stance and the smallest offset (1.8 mm) for the widest stance. There was also a reliably ($F(1,9) = 58.80$; $P < 0.01$) greater offset on push (2.5 mm) compared to release (1.7 mm).

Discussion

We have described the effect of moderate sideways forces applied to the hip. The form of the hip displacement function immediately after force onset (push) is a relatively rapid movement in the direction of the applied force. This lateral sway is arrested and hip position then reverts, with a little oscillation, to a value close to that before force onset. On removal of the sideways force (release), hip position swings over in the other direction (opposite to that of the previously applied force) before returning to an offset close to that at the beginning of the trial. The results of the two experiments show that sway increases with force and decreases with stance width. The peak displacement and the return time are less on release than on push.

Sideways sway of the pelvis relative to a fixed base of support is presumably resisted by the hip joint abductors of the leg towards which the sway is directed, aided by the hip joint adductors of the other leg. With larger applied force, there is greater acceleration of the body mass. If the muscles opposing the force are activated after a fixed latency, greater hip displacement would be expected before muscle-induced deceleration. This would explain the greater peak displacement with greater force. It is also possible that the muscle force used to decelerate the hips does not increase in proportion to the applied force. If so, this would also contribute to increase in peak displacement with applied force. However, one argument against this interpretation is that the time taken to return the hips to their initial position against the applied force did not increase reliably with force level. This implies that muscle force levels were raised in proportion to the applied force.

We now turn to consider the different effects of push and release. During quiet standing prior to push, little activity is required of the hip ab/adductors to maintain hip position in the frontal plane. One might therefore speculate that the initial response to push is reciprocal activation of the bilateral ab/adductor pairs, culminating in a steady level of activation in the ab/adductor pair (say, A) towards which the force continues to be directed. It may be supposed that it is this persisting muscle activity in A that, on release, results in the sway directed back towards the source of the original applied force. There would then appear to be at least two options to restore hip position. The continuing activity in A might first be inhibited, then the opposing ab/adductor pair (B) would be activated (reciprocal activation). Alternatively, perhaps to avoid relatively large lags in inhibiting force developed by A, B might be coactivated immediately release is detected. In the latter case, the effective stiffness of the system would be higher than for reciprocal activation. In this regard it is interesting to note that the smaller peak hip displacements and shorter return times seen on release than on push are consistent with higher effective stiffness. A fruitful direction for future research would therefore be an electromyographic investigation of the activity patterns of the hip ab/adductors to determine their degree of coactivation on push and release.

In the second experiment we evaluated the effects of sideways forces as a function of stance width. We found stability in the frontal plane increased as the feet were

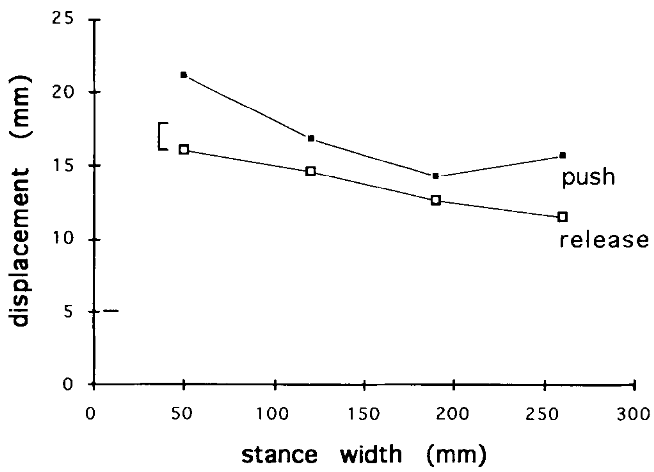

Figure 4. Experiment II. Decrease in peak hip displacement with stance width on lateral force push and release. Group mean data (with two standard errors) for 10 subjects.

placed further apart. This result extends the findings of Day et al.[14], who showed that natural body sway in the frontal plane (standard deviation of hip or shoulder position or standard deviation of lateral centre of pressure) decreased with stance width. One interpretation of the effect of stance width is that there is more effective force generation by postural muscles when the feet are further apart. Another possible interpretation is that there is greater sensitivity to sway with feet apart. This might, in turn, allow earlier, and so more effective, corrective muscle action. Again an electromyographic investigation would be of interest.

In the experiments reported in this paper the subjects were healthy adults. Previously we have reported the effect of lateral pushes at the hips on stroke patients[13]. Not surprisingly, stroke patients tended to sway more, both on push and release. More useful for documenting the asymmetry of motor function after stroke, the method revealed a clear asymmetry in peak hip displacement and return time depending on whether the applied force was directed towards the involved, hemiparetic side or the non-involved side. The peak displacement was greater and it took longer to restore hip position when the sway was towards the involved side than towards the non- involved side. Interestingly, the difference between non-involved and involved sides was more pronounced on release than on push. The method thus appears to have utility in clinical evaluation as well as fundamental research.

In conclusion we have described a method for testing standing stability. We have demonstrated its use in the frontal plane and documented increases in sway with increased force and with narrower stance. We have shown that the application of force causes more disturbance than its termination. We consider that the technique offers an interesting alternative to displacing or rotating the base of support in the assessment of the neural control of posture and balance. Because the method involves a mechanical link at the waist it may be particularly appropriate in the evaluation of patients with impaired control of equilibrium[15]. With hybrid control over position as well as force, it should be possible to stabilize or restore hip position over the base of support when a perturbation would otherwise have led to a fall. Further development of the present approach along these lines might then offer a very real potential not only for assessment but also for an innovative exercise machine for retraining the retaining of balance.

Acknowledgements

This research was partially supported by a grant from the Chest, Heart and Stroke Association.

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