A POSTURAL MODEL OF BALANCE-CORRECTING MOVEMENT STRATEGIES

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Abstract—The patterns of joint torques and movement strategies underlying human balance corrections were examined using a postural model. Two types of support-surface perturbation, dorsiflexion rotation (ROT) and rearward translation (TRANS), were employed. These two perturbations were adjusted to produce similar profiles of ankle dorsiflexion in order to obtain information on the role of lower leg proprioceptive inputs on triggering balance corrections. In addition, the dependence of balance control on head angular and linear accelerations was investigated by comparing the responses of normal and vestibularly deficient subjects under eyes-closed and eyes-open conditions. Differences in ROT and TRANS movement strategies were examined in three ways First, the amplitude and polarity of active joint torques were analysed. These were obtained by altering joint torques applied to a postural model until movements of the model accurately duplicated those of measured responses. Second, the pattern of body-segment angular movements depicted by stick figures moving in response to the computed joint torques was investigated. Third, the peak amplitude and patterns of crosscorrelations between joint torques were measured. Active ankle, knee, and hip joint torques computed for normal subjects rotated the body forward for ROT. In the case of TRANS, computed active torques in normals were of opposite polarity to those of ROT and reversed the forward motion of the body. Subjects with vestibular deficits had lower amplitude torques for ROT and failed to counter the platform rotation. Hip torques for TRANS in vestibular deficient subjects were of opposite polarity to those of normal subjects and resulted in excessive forward trunk rotation. Normally, neck torques acted to stabilize the head in space when trunk angular velocity peaked. Vestibular deficient subjects displayed head movements in response to ROT similar to those generated when neck torques were absent. For TRANS, these same subjects exhibited overcompensatory neck torques. Stick figures of normal responses indicated a stiffening of the body into a leg and a trunk-head link for ROT and a flexible multilink motion for TRANS. Likewise, normal response strategies, defined by using crosscorrelations of joint torques, differed for ROT and TRANS. All joint torque crosscorrelations were significant for TRANS. Neck torques led those of all other joint torques by 40 ms or more, and hip joint led ankle torques by 30 ms. Joint torque correlations for ROT were organised around hip and ankle torques without a major correlation to neck torques. Fundamental changes in all torque crosscorrelations occurred for vestibularly deficient subjects under both eyes-open and eyes-closed conditions. These results support the hypothesis that the modulation of postural responses by vestibular signals occurs at all major joint links of the upright human body and that the strategy underlying balance corrections at the hip and neck is selected independent of local sensory input from the lower leg. Rearward translation and dorsiflexion rotation of a support-surface elicit different movement strategies when ankle angle changes are matched for such disturbances to human upright balance.

Keywords—human postural control; models of postural control; vestibulo-spinal reflexes; muscle stretch reflexes; peripheral vestibular deficit.

Introduction

Natural human movements, including balance corrections, may be organised as if the head,
and to a lesser extent the trunk, adopted a stable posture around which the arms and legs move, behaving like the pneumatic pistons of a robot system (1). If the head is accurately stabilized in space during balance corrections, the activity of neck and trunk muscles should be closely related to each other as well as to the activity of leg muscles. The resulting active joint torques at the hip and neck would necessarily covary strongly with those of the ankle and knee in order to produce a stable posture correction. In other words, the pattern of the joint torque correlations could completely define the movement strategy because the angular displacements of each body segment are uniquely dependent on the pattern of applied joint torques.

To date, movement strategies have been defined in terms of the joint about which most rotation appears to occur. For example, a hip strategy was described by Horak and Nashner (2) as a complex coordination of hip and leg muscles that rapidly rotates the trunk forward and the legs backwards in order to translate the body's centre of gravity rearwards. The elicited this response pattern when subjects stood on a narrow beam that translated backwards. The ankle strategy, consisting predominantly of activity in leg muscles, was first observed as the response to slow translations (about 15 cm/s) when subjects stood normally. This concept was extended to describe responses to larger and faster (about 30 cm/s) translations (3). Later, the strategy observed for fast translations was redefined as a "mixed" or "multilink" strategy (4,5). The strategies elicited by different speeds of support-surface rotation have been explored less extensively. Generally, it appears that a stiffening of body links occurs that holds the body upright against a tilted support-surface (6,7).

The hip, ankle, and stiffening strategies may influence the task of stabilizing the head in space quite differently. The greatest neck movement, relative to the trunk, should occur with the hip strategy, and the least with the stiffening strategy, because the trunk and head move as a unit. If head movements are to be included in the movement strategy associated with support-surface translation, the resulting balance corrections will become more multilink in character. In summary, the covariation between ankle, knee, hip, and neck torques should be different for a stiffening strategy in response to dorsiflexion rotation compared to a multilink or mixed strategy in response to rearward support-surface translation.

To establish a unifying concept describing different movement strategies during balance corrections, the model of postural control, presented in this paper, was developed. The model accurately matches leg, trunk, and head movements to those induced by support-surface translations and rotations. By cross-correlating the active joint torques required to fit the model to experimental data, fundamental differences were established in the pattern of movement strategies required across all joints to correct these two types of balance perturbations. The amount of ankle dorsiflexion for both types of balance perturbation were matched. In addition, responses of normal and vestibularly deficient subjects were compared. Using both techniques it was possible to provide some insights into the types of sensory inputs triggering the selection of a particular movement strategy.

The question of the sensory inputs required to trigger and modulate human movement strategies has remained unresolved. Macpherson and colleagues (8) demonstrated in cat, that the movement strategy in response to translation and rotation were identical provided the stimuli were amplitude controlled to produce similar proprioceptive inputs at the foot. On the basis of this finding, it was argued that the balance-correcting movement strategy was triggered by proprioceptive inputs from the cat's feet. These authors also argued that vestibular inputs could not lead to an identical response strategy for both types of perturbation because vestibular sensory inputs were probably very different for rotation and translation. Such results suggest a predominance of proprioceptive inputs underlying balance corrections in cats. Two important research findings suggest a different
sensory predominance in humans: a) the inability of subjects with bilateral vestibular deficit to maintain an upright stance following rotation of the support surface under eyes-closed conditions (6), and b) the inability of subjects with the same deficit to perform the hip strategy in response to translation (9). Because of the apparent discrepancy between human and cat movement strategies, we chose to repeat the experiments of Macpherson and colleagues (8) using the same experimental protocol in humans. The experimental data described here have been presented in preliminary form elsewhere (5).

**Methods**

Balance corrections were induced in subjects standing upright, using either a 4-cm (21 cm/s) rearward translation (TRANS) of the support surface or a 3.4° (27°/s) dorsiflexion rotation (ROT). This choice of stimulus parameters yields ankle dorsiflexion trajectories that were similar over the first 200 ms, but, as Figure 1 illustrates, head angular and linear acceleration profiles that were opposite in polarity. Techniques employed to record the biomechanical and electromyographic traces shown in the figures have been described elsewhere in detail (5,6,10). The main difference between this and previous studies (6) was the degree of body lean that subjects maintained until the onset of the support-surface movement. In this study, the sum of the ankle torques exerted on the support surface under each foot was controlled to within 1 Nm. The ankle torque obtained when subjects stood in their normal standing position, whether it was at exact upright or not. Subjects could control their leaning via a visual display of pitching ankle torque (equivalent to centre of foot pressure) at eye level 1 m from the subject and via tone signals. As in the previous studies, the following measurements were recorded:

- head angular velocity and acceleration in the pitch plane;
- head anterior–posterior linear acceleration;
- trunk angular velocity in the pitch plane;
- shank pitch angle (right lower leg) with respect to the vertical;
- torque about the ankle joint of the right foot;
- support-surface rotation and translation.

Torque about the right ankle joint of "ankle torque" was computed from the equivalent moment generated by the ground reaction forces. These were measured with strain gauge systems located at the corners of the support surface under the right foot, and, for ankle torque calculations, referred to an axis colinear with the axis of the ankle joint. The right shank angle with respect to vertical was measured with a potentiometer system built into the axis of platform rotation. This axis is colinear with the axis of ankle rotation. A vertical bar was attached to the slider of the potentiometer. A hinge at the bar's attachment to the slider permitted lateral movements of the bar without changing the amount of bar rotation about the potentiometer axis. The upper end of the vertical bar was attached to the lateral aspect of the right leg using a Velcro® strap placed around the right calf 10 cm below the knee.

EMG recordings were taken from the tibialis anterior (TA), soleus (SOL), rectus femoris (quadriceps – QUAD), biceps femoris (hamstrings – HAM), lumbar paraspinalis (PARAS), rectus abdominis (ABDOMS), and trapezius (TRAP) muscles.

The balance perturbations consisted of a series of 11 rotations followed by 11 translations. These stimuli were presented first under eyes-open conditions, then were repeated under eyes-closed conditions. Off-line, individual response averages were computed for the first 3 and last 8 of a series of 11 trials. The response of the last 8 trials was the concern of the present study because adaptation effects present in the first 3 trials were not of interest (compare to 6). To facilitate comparisons between populations, across-subject averaging was performed on all 7 normal subjects and 2 subjects with bilateral vestibular deficit (no...
Dorsiflexion Support-Surface Rotation

Figure 1. Profiles of head accelerations and ankle dorsiflexion elicited by a dorsiflexion of the support-surface (upper traces) and translation of the support-surface (lower traces). Each biomechanical trace is the average of 56 responses from 7 normal subjects: the 8 responses per subject are the last 8 of a series of 11 stimulus presentations. Full scale deflection of the traces is indicated at the upper and lower borders of each frame. Upward deflection of angular movement traces indicates backward pitching in this and all succeeding figures. The head pitch angular acceleration has a different scale for translation. Backwards linear acceleration of the head is represented by a downward deflection of the trace. Note that the linear acceleration transducer mounted on the helmet worn by the subject also registers changes in head angular-accelerations (tangential component) and head tilt. Initial ankle dorsiflexion is similar for the two perturbations. Head accelerations are of opposite polarity for rotation and translation of the support-surface.
caloric response, no response to 80°/s² acceleration steps during horizontal vestibulo-ocular reflex rotating-chair tests).

The biomechanical responses were compared with those of a 4-segment (shank, thigh, trunk, head) computer model of human posture (see Figures 2 and 3). The model consisted of the equations-of-motion of the 4-segment linkage with simple hinge joints in the sagittal plane whose active and passive torques could be varied by the experimenter. The effects of this variation were displayed on the computer console as a deviation from the average population data being modelled (see plots in the lower-left panels of figures 2 and 3).

The equations-of-motion were similar to those employed by Koozekanani and colleagues (11,12). The head was a separate, fourth link. Motion of the model was driven by the support-surface movement and by active and passive torques at the hinges. Passive torques were dependent on the masses and moments of inertia of each segment and the spring and dashpot constraints at each joint.

Active model torques could be applied at each link by the experimenter. Masses and moments of inertia for each segment were equated to those reported by Dempster (13) after correction for the average height and weight of each population. The arms were considered to be part of the trunk segment. The spring and dashpot constraints were assumed to be equivalent to the passive and active (due to background muscle activity) visco-elastic resistance at the ankle and other joints. The values of these constraints were estimated by varying the coefficients of each joint's visco-elasticity until a best fit between model and average recorded data was obtained for the first 70 milliseconds of the responses. Initial values for the ankle joint visco-elasticity were collated from the available literature (14,15). The 70-ms limit was based on the time when stretch-reflex responses of ankle muscles can first be observed in ankle torque responses (14). Once estimated, it was assumed that visco-elastic parameter values of the model remained constant throughout the balance perturbation and, with the exception of the knee joint, that additional changes in biomechanical responses different from those of the mass-spring-dashpot system would be due to active torques applied at each of the segment joints.

To model the knee-joint locking mechanism beyond 179° of knee extension, passive visco-elasticity was increased until, at 185° of extension, it had increased 10-fold.

An iterative approach was used to determine the successful active joint-torque combinations yielding a best fit between model and experimental data for 70 to 600 ms. Normally, ankle torque response was fitted. Then, in 100- to 150-ms time segments, knee and hip torques were adjusted simultaneously until lower leg angle and trunk angular velocity responses were accurately modelled. Readjustment of model ankle torque was necessary once knee and hip torques were approximately modelled. Finally, the neck and hip torques were varied until both model and experimental data for head and trunk angular velocity were concurrent. This modelling approach was extremely time consuming and, therefore, was limited to population rather than individual responses.

Each model torque could consist of a sum of up to 8 cosine-bell-on-a-pedestal torques whose onset, polarity, rise and fall time and amplitude could be specified by the user. A cosine-bell-on-a-pedestal is similar to the time function used to taper-window time series prior to fast Fourier transformation (see Figure 9.13 of reference 16). Its leading and trailing edges are defined by one cycle of the function \(1 - \cos \omega t\), where \(\cos \omega t\) starts with the value \(-1\) and ends with the value \(+1\). The angular frequency, \(\omega\), chosen determines the rise or fall time of the time function.

All simulations were performed with Euler-Cauchy integration (17) of the model's equations-of-motion. The final simulation was rechecked for accuracy with Runge-Kutta-Fehlberg integration (17,18). No differences were observed between the final Euler-Cauchy simulation and that performed with Runge-Kutta-Fehlberg integration.

The specific solution sought in these simulations was, given a specific balance distur-
Figure 2. Average response patterns of a group of 7 subjects with normal balance to dorsiflexion rotation of the support-surface when tested under eyes-closed conditions. The average surface EMG responses are shown (C). With the exception of trapezius, muscles on the ventral surface of the body are plotted positive downwards; dorsal muscles, positive upwards. The deflection amplitude of one vertical gradation is noted next to the muscle name. Average biomechanical responses are shown (B). The experimental data are shown as open circles and the model responses by full or interrupted lines. Backward pitching of the lower leg with respect to vertical, and trunk and head angular velocity are shown by an upward deflection of the traces. Plantar-flexion–ankle torque imposed on the support-surface is shown by upward trace deflection. Full-scale vertical deflection (panel edges) of each trace is indicated at the upper and lower borders of each frame. Active torques required at each joint to produce the model responses are shown (D). Upward deflection of torque traces indicates rearward pitching of body segments above the joint when body segments below the joint are fixed. The trunk angular velocity (model and experimental data) and platform angle traces are repeated for clarity (A). The model movements are shown as stick figures at every 50 ms from onset of platform movement. The angular velocity of each segment is shown as an arrow. (Figure continues)
Balance—ROT or TRANS—define a feasible set of joint torque combinations that would restore balance in a way identical to that with which normal or vestibularly deficient subjects restore, or attempt to restore, balance. No efforts were made to determine the uniqueness of the solution. Solutions presumably had a well-defined minimum because even small 1-Nm variations in knee and hip torque profiles of the best-fit model yielded evident discrepancies between model and real data.

Movement strategies were defined on the basis of normalized crosscorrelations or covariance functions (equation 3.54 of reference 16) between the computed active torques and on the basis of the angular movements of the model stick figure (D and A of Figures 2 and 3, respectively). The peak positive amplitude
Normal

Rearward Support-Surface Translation

Eyes Closed

Figure 3. Average response patterns of a group of 7 subjects with normal balance to rearward translation of the support-surface when tested under eyes-closed conditions. For details, see legend to Figure 2. (Figure continues)

(with respect to the correlation due to the mean torque level between 0 and 600 ms) and its latency were used as descriptors of the normalized crosscorrelations. Significance measures of crosscorrelation peaks remain an unresolved statistical issue (19). For this reason, we arbitrarily chose a correlation coefficient 0.5 above the mean level as significant.