Strategies for maintaining a handstand in the anterior-posterior direction

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ABSTRACT

KERWIN, D. G., and G. TREWARTHA. Strategies for maintaining a handstand in the anterior-posterior direction. *Med. Sci. Sports Exerc.*, Vol. 33, No. 7, 2001, pp. 1182–1188. **Purpose:** The purpose of this analysis was to determine the contributions made by wrist, shoulder, and hip joint torques in maintaining a handstand. **Methods:** Handstand balances ($N = 6$) executed on a force plate and recorded with two genlocked video cameras were subjected to inverse dynamics analysis to determine anterior-posterior joint torques at the wrists, shoulders, and hips. Multiple regression analyses were conducted to investigate which of the joint torques were influential in accounting for anterior-posterior whole-body mass center (CM) movement. **Results:** Results demonstrated that, in general, all calculated joint torques contributed to CM movement. In a number of trials, wrist torque played a dominant role in accounting for CM variance. Ostensibly, superior handstand balances are characterized by important contributions from wrist torques and shoulder torques with little influence from hip torques. In contrast, hip torques were found to be increasingly influential in less successful balances. **Conclusions:** It is concluded that multiple joints are utilized in maintaining a handstand balance in the anterior-posterior direction, and there appears to be two joint involvement strategies, which supports similar findings from postural research on normal upright stance.
Key Words: BIOMECHANICS, GYMNASTICS, INVERTED BALANCE, JOINT TORQUES

The maintenance of balance is fundamental to all postural activities, requiring a coordinated response by the central nervous system (CNS) to information obtained through the proprioceptive, visual, and vestibular feedback systems. Postural dynamics are complex, not only concerned with the maintenance of relative positions of body segments but also with fine adjustments (11), where distinct control strategies exist for all movements to ensure that balance is preserved. Previous research has indicated that altered body orientation may influence selected balance strategies through differences in body-surface contact area (4), strength of supporting segments (8), and disrupted environmental information (12).

The maintenance of upright stance has received considerable research attention. Studies have generally employed EMG to record muscle activation, or inverted pendulum models, as means of delineating existing balance strategies. Generally, these research approaches have achieved similar findings with two main strategies being identified: an “ankle strategy” and a “hip strategy.” In the ankle strategy, there is a preference to rotate predominantly about the ankle joint with the knee and hip joints remaining relatively immobile. The hip strategy involves flexion/extension of the hip joint, which produces an inertially driven torque that acts at the body-support surface interface. It has been reported that the choice of strategy is determined by the amount of CM movement relative to the center of pressure (CP) excursion limits (13). When the body CM moves within the CP excursion limits, the ankle strategy is utilized. However, during larger CM disturbances, which exceed the biomechanical constraints allowing adjustment potential at the ankle, individuals are seen to switch to the hip strategy.

Despite the conjecture that altered body orientation produces differences in balance strategies, inverted stance strategies have received scant attention. Nevertheless, an examination of CP excursions in divers performing a number of straight body postures found that trials with smaller support surface areas, particularly those involving inverted stances, were associated with greater shifts in CP position (10). These greater CP excursions in inverted postures were thought to reflect a reduced potential to generate the requisite moments about the wrist as compared with the ankle in upright stance or additionally result from altered body mass distribution from the axis of oscillation. Furthermore, Slobounov and Newell (12) identified qualitative compensatory movement strategies utilized by individuals executing inverted stance and found one of the most common strategies to be anterior-posterior oscillations of the feet. It was concluded that movements about the ankle joint would provide the initial tool for maintaining balance. A kinematic

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analysis was also conducted on the data showing that increases in the body's range of motion throughout a trial occurred primarily in distal body segments, the lower leg and about the elbow joint, whereas the head and trunk remained relatively fixed.

Resulting from these examinations, it is apparent there are several possibilities that may exist for balance mechanisms used in maintaining inverted stance. Results from postural research have shown that for a well-learned "automatic" activity, such as normal stance, large perturbations of the support surface are required before a hip strategy is employed. Inverted stance is a less well-practiced activity that is also inherently less stable due to a smaller base of support and less ability to produce torques about the wrist as compared with the ankle. Consequently, it seems reasonable to suggest that because of the relatively unlearned nature and inherent instability of the skill that during inverted stance hip torques may be employed at the extremes of support even without the presence of surface perturbations or external disturbances. Therefore, it is possible that inverted stance strategies may be similar to strategies used in normal stance with external perturbations, that is with the joints close to the support surface acting as primary contributors to maintaining controlled balance and with joints further from the support surface only being utilized at the extremes of balance. It has been noted that individuals may struggle to exert adequate torques about the wrist joint to maintain balance and also that movements about joints far from the support surface, such as the ankle, may be employed as balancing mechanisms. Nevertheless, from a gymnastics viewpoint, "good body form" dictates that the handstand balance is executed with straight arms and plantar flexed ankles. Therefore, the potential influence that certain body joints (elbow and ankle) can contribute to the maintenance of balance may be restricted.

The purpose of this analysis was therefore to address two related research questions: Which joints are used in maintaining a handstand balance and are joint torques used in a strategic manner?

METHODS

Data collection. University Ethics Committee approval was obtained for a study of six male gymnasts to perform a series of handstand balances under laboratory conditions. Individual informed consent was given either by the subject or for those under the age of 18 yr by their accompanying guardian. Subjects' mean age, mass, and height were 15.7 $\pm$ 2.7 yr, 53.63 $\pm$ 10.04 kg, and 1.62 $\pm$ 0.08 m, respectively, and all were of national junior squad standard. A calibration structure comprising six upright poles, each with five markers at vertical intervals of 0.5 m, was positioned around a force plate (Kistler 9281-B12, Switzerland) forming a rectangular base of 1.025 m $\times$ 0.910 m. Images of the calibration structure were recorded before the subject trials. Force and video data were recorded for each gymnast executing a series of three handstand balances of 5-s duration, one of which was selected for further analysis. Two video camera recorder systems (Sony Hi8 Handycam CCD and Sony Hi8 Hyper HAD 3CCD, Japan) were used. The genlocked cameras were located on either side of the force plate viewing the gymnast in handstand from the rear. The optical axes of each lens made an angle of 45° to a center line running along the mid-line of the force plate. The cameras were operated at 50 fields per second with electronic shutters set to 1/215 s and 1/250 s, respectively. The force plate data were sampled via an analog to digital converter (CED1401, Cambridge Electronic Design, United Kingdom) at 200 Hz. Force and video data were synchronized using a pair of custom built linear LED arrays, each of

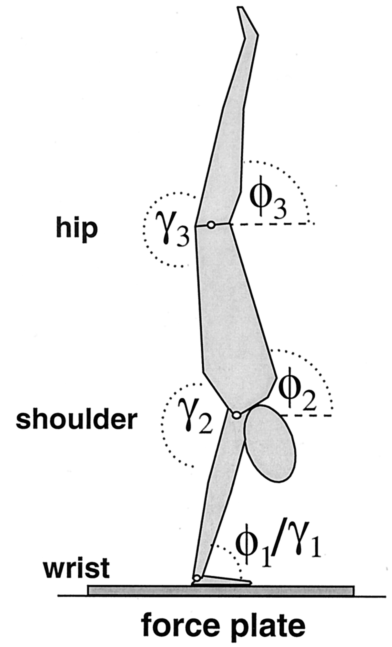

FIGURE 1—Definition of segment angular position (ϕ) and joint angles (γ) at the wrist ($_1$), shoulder ($_2$) and hip ($_3$).

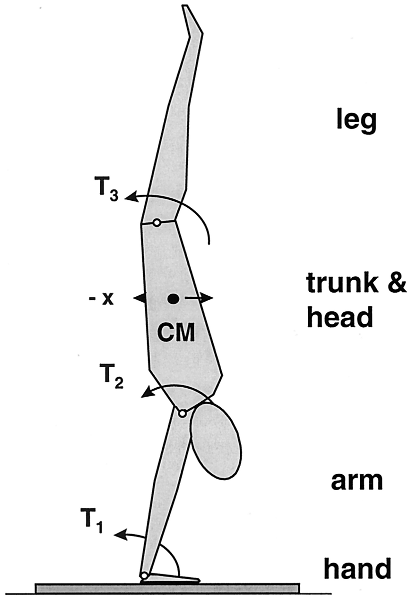

FIGURE 2—Four-segment model of the gymnast showing mass center (CM) and torques at the wrist (T_1), shoulder (T_2), and hip (T_3).

which had been positioned within the field of view of the cameras. The force capture and LED synchronization unit were triggered simultaneously at the start of the data recording period. The 20 LEDs were illuminated sequentially at 1-ms intervals. Half a second later, all the LEDs were extinguished. By observing the number of illuminated LEDs in the unique initial video field for each view, the time offset in milliseconds from the start of the force capture could be determined. Superficial circular body markers were positioned on the posterior surface of each subject at points corresponding to selected joint centers. These markers were subsequently digitized throughout a selected 5-s handstand trial for each subject. Customized body segment inertia parameters were generated from anthropometric measurements by using the inertia model (17), which also included a mass ratio correction factor to ensure that the calculated and directly measured body mass values corresponded.

Data processing. Video data were digitized using the TARGET high-resolution system, developed at Loughborough University as an improved version of the Prisma III system (6). Images of the calibration structure were digitized for each camera view. Camera calibration was achieved using an 11-parameter Direct Linear Transformation procedure (5), and unbiased estimates of reconstruction accuracy were determined. The two views of each handstand were digitized. In each field, 17 body landmarks corresponding to fingers, wrists, elbows, shoulders, hips, knees, ankles, toes, the center of the head, and two reference points were digitized. The three-dimensional real-space coordinates of the joint centers were reconstructed and reliability estimates obtained. Interpolating quintic splines (15) were fitted to the digitized coordinates to facilitate the estimation of joint center locations at times in between those corresponding to video fields. The LED arrays were used to calculate a time offset between the trigger of force data capture and the first video field. This offset was applied to determine the point at which interpolated coordinates should be synchronized with the force data, thereby matching the two data sets to within 1 ms. Matched force and video data were output at 50 Hz.

Force-synchronized coordinate and inertia data were combined to determine values for selected segment angles, joint center coordinates, and whole-body CM coordinates. A pseudo coordinate data set was obtained from the interpolated coordinate data set by replacing every alternate data point with the point representing the average between the preceding data point and the proceeding data point. The two coordinates data sets and two sets of repeated digitizations were used in combination to generate four sets of angle estimates. A quintic spline was fitted through the mean of the angle estimates with tightness of fit determined using the method of Yeadon (16). Angular velocities and angular accelerations at each joint were obtained from the splined angle values. Splined angles were also manipulated geometrically to obtain the wrist, shoulder, and hip joint angles as illustrated in Figure 1.

Coordinate and angle reliability estimates were calculated by generating pseudo data sets, for comparison, by averaging data from adjacent times and again using the procedure of Yeadon (16). CM position values were treated for the systematic offset introduced by digitization of markers on the back of the body. Horizontal and vertical displacement offsets at each estimated joint center were determined from direct subject measurements. An additional correction was applied to the body CM horizontal position by minimizing the RMS difference between CM and CP position data so that although the CM generally moved within the base of support (CP) the mean values of the two time histories were equal (3).

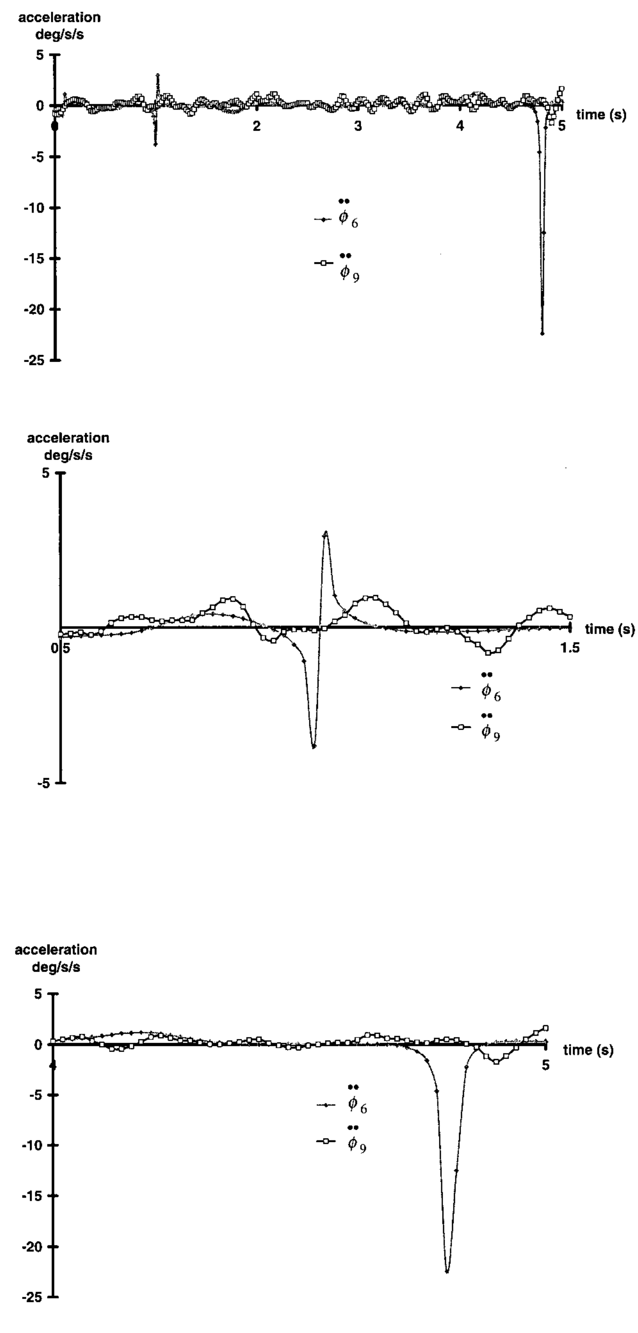

FIGURE 3—Outlying acceleration values when solving from six equations ($\ddot{\phi}6$), and the adopted solution of using nine equations ($\ddot{\phi}9$) to solve for three unknown joint torques (times 0.5–1.5 s and 4.0–5.0 s are shown expanded to highlight the "outliers").

Inverse dynamics. For the present study, the focus was on joint strategies for balance in the anterior-posterior direction. The amount of rotation about a longitudinal axis, defined as a line joining the mid-points of the knees to the mid-points of the shoulders, was less than 0.5° for all subjects, and the amount of CM movement (as a proportion of the base of support) in the medio-lateral direction was only 20% of the movement in the anterior-posterior direction. Therefore, for this study, all inverse dynamics calculations proceeded in two dimensions along a sagittal plane. Each gymnast was represented in the xz-plane by averaging the left and right joint center coordinates and treated as a four segment model (Fig. 2), comprising a hand, arm, trunk and head, and leg segment, with three corresponding joints: wrist, shoulder, and hip.

The data were subjected to inverse dynamics analysis. For each of the four segments, three equations of motion were developed: one for resultant vertical force, one for resultant horizontal force, and one for moments about the mass center. Eliminating reactions at each of the three joints left six equations to solve for six unknowns: wrist, shoulder, and hip torques (T_1, T_2, T_3) and angular accelerations ($\ddot{\phi}_1$, $\ddot{\phi}_2$, $\ddot{\phi}_3$), respectively. Figure 3 shows two sample plots of wrist angular acceleration for trial B1. One plot is based on the six-equation solution and highlights several "outlying" extreme accelerations. Three further equations were added based on the acceleration estimates obtained from the quintic spline fit of the angle data. This resulted in nine equations that could be solved in a linear least squares manner to obtain the three required joint torques and the three angular accelerations without the presence of extreme "outlying" values. All calculated joint torques were subsequently normalized. Individual joint torques were divided by the subject weight and height (9) and multiplied by the group mean weight and height. This normalization procedure maintained values in the same SI units throughout and accounted for the fact that individuals with greater masses and heights required greater torque to be exerted for a given displacement of the CM.

Data analysis. Slobounov and Newell (12) found that increases in the body's range of motion during a handstand trial occurred primarily in distal body segments and furthermore observed a reduced coupling between postural movements during a trial. These two effects were indicative of pursuing instability; thus, some form of quantification of these phenomena may provide a measurement of "steadiness" of balance. For instance, calculation of horizontal excursions of the ankle joints about a certain mean may represent the body's range of motion. An alternative form of classifying steadiness or solidity of balance may be an examination of the variance in torque values over time.

Maintaining balance is fundamentally a task of preserving the CM position in dynamic equilibrium within the boundary limits of the base of support. Individuals control the CM position by exerting torques about joints to move segments into body configurations that will facilitate stability. Hence, an examination of the joint torques that are responsible for causing CM movement may directly lead to an appreciation of the balance strategies utilized by gymnasts in handstands.

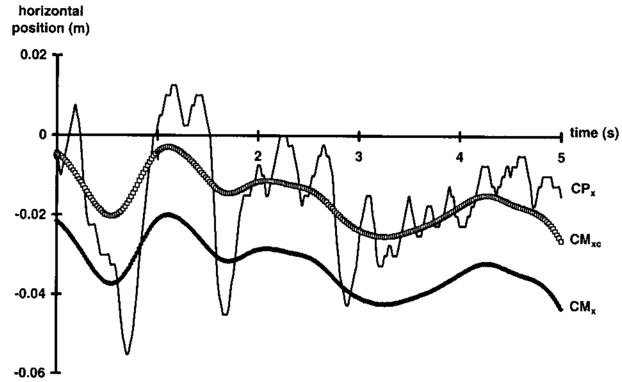

FIGURE 4—Comparison of CP_x (horizontal CP motion) to CM_x and CM_{xc} (horizontal CM motion) before and after final systematic correction.

Accordingly, applying the values obtained from the inverse dynamic analysis, CM positions were regressed against T_1, T_2, and T_3 by using the forward stepwise estimation multiple regression method (SPSS® Base 7.0 for Windows™, 1996). These regression calculations determined the amount of CM position variance that could be accounted for by a collaboration of wrist, shoulder, and hip torques.

RESULTS

The reconstruction accuracy of the calibration markers was 0.0033 m, 0.0027 m, and 0.0022 m in the anterior-posterior, medio-lateral, and vertical directions, respectively. Average reliability estimates of the reconstruction of individual body landmarks ranged from 0.008 m to 0.009 m. The averaging effect of using all 17 data points, each with random error, in the calculation of CM position improved the reliability for estimating the location of the whole-body CM by a factor of $\sqrt{17}$. The resulting reliability values were within the expected range of between 0.0016 m and 0.0020 m. It is recognized that the inertial model of the human body used in this study was originally employed to calculate the CM location of airborne humans, a case in which the gravitational effect on the internal viscera of the body can be neglected. During a handstand, however, gravity exerts a compressive influence on body viscera causing the CM to be located closer to the hands (base of support) than would be

TABLE 1. Measures of steadiness: ranking of gymnasts based on joint torque variation and ankle excursion.

	T_1(det) (N·m)		T_2(det) (N·m)		T_3(det) (N·m)		x_{ankle} (m)		
Trial	SD	Rank	SD	Rank	SD	Rank	SD	Rank	Rank
B1	6.081	1	4.010	2	2.878	1	0.015	1	1
B2	6.206	2	4.261	4	5.891	5	0.020	2	3=
B3	7.228	4	4.133	3	2.983	2	0.035	4	3=
B4	8.875	6	4.458	5	6.751	6	0.094	6	6
B5	6.289	3	3.961	1	4.253	3	0.022	3	2
B6	7.246	5	4.883	6	4.494	4	0.047	5	5

T_1, wrist torque; T_2, shoulder torque, T_3, hip torque; (det), detrended torque data; x_{ankle}, horizontal excursion of the ankle joint center.

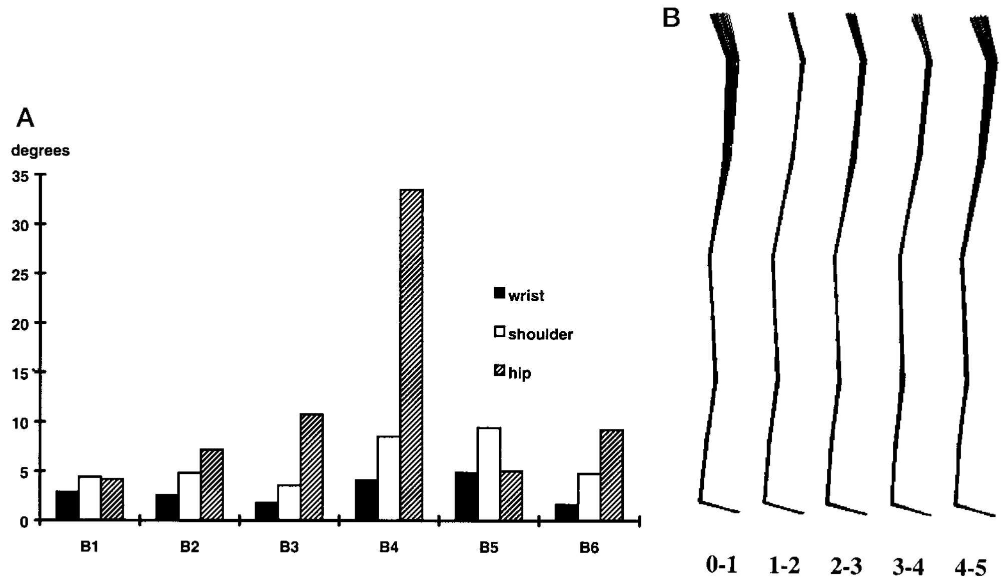

FIGURE 5—A, Comparison of joint angle ranges between trials. B, Stick figure graphics showing increasing ranges of motion arising in the distal segments for five 1-s time intervals for subject B1.

the case when the body was airborne. Nevertheless, it is estimated that this CM shift would have less than a 2% influence on total sway path during a handstand trial and that this influence would be of a systematic nature and would affect all subjects equally. The reliability estimates for angle values ranged from 0.9° to 2.1° across trials. Body surface marker coordinates were adjusted to allow for the differences between the markers and the directly estimated joint center locations. A further horizontal offset correction (0.0152 m $\pm$ 0.0080 m) was applied to bring the mean CM position onto the mean CP position, shown in Figure 4.

Two measures of "steadiness" were employed to rank the gymnasts in balancing ability. These were standard deviations of detrended torque values (reflecting variability of torque time histories about a mean of zero) and standard deviations of horizontal ankle joint center excursions. The first measure provided an indication of the amount that joint torque had to vary over time to control CM position, and the second indicated the amount of overall body motion. Values for each trial are shown in Table 1. Similar ranking occurred for each gymnast with both scoring systems, and thus an overall ranking based on the individual scores was employed (final column of Table 1).

The range of motion exhibited at each joint was determined by an investigation of joint angle ranges. In general, the range of motion was smallest at the wrist, larger at the shoulder, and greatest at the hip, although some variability in results existed between trials as shown in Figure 5A. These results are further illustrated in Figure 5B. Each image represents a cluster of stick figures at 0.02-s intervals for five 1-s periods. Small amplitude oscillations can be seen for the arm segment. These small oscillations at the lowest segments are amplified as segments further from the base of support are considered. This effect is particularly evident in images corresponding to "0–1 s" and "4–5 s." The graphics sequences represent subject B1, who was ranked number one and thus has been regarded as a "stable" hand balancer.

Torque-time histories obtained from two-dimensional inverse dynamic analysis for wrist, shoulder, and hip joints were similar in many respects across trials. Wrist torque exhibited the greatest magnitude in all trials, with a positive torque tending to increase the wrist joint angle. Shoulder torques were generally of a lesser magnitude and could be either positive or negative depending on the trial. Hip torques had the smallest magnitude and again could be either positive or negative in direction. Figure 6 provides examples of torque-time graphs obtained in trial B5.

Forward stepwise estimation regressions carried out on CM position as the dependent variable against joint torques as independent variables for 251 readings per subject determined which joint torques were prominently related to CM movement (1). Table 2 contains the results of these regressions and includes total accounted variance (adjusted), stepwise order of inclusion, and the levels of significance for the independent variables in the final regression model.

DISCUSSION

In general, all three identified joint torques contributed to CM movement. In a number of trials, wrist torque had the most dominant role in accounting for CM variance, followed in order by shoulder torque and hip torque. The fact that wrists have been identified as the most influential joint in maintaining CM position appears to be congruent with the views of many gymnastic texts, for example the "on-line balancing" technique (2), where the wrists act as the "control centers." Furthermore, in terms of "gymnastic form," it is necessary that gymnasts hold a rigid, straight position. Therefore, the generation of torques about the point of contact rather than elsewhere in the link system would provide efficient control and minimize movement in the upper segments.

Slobounov and Newell (12) found that one of the primary balancing strategies for inverted stance was anterior-posterior oscillations of the feet. Examination of present data

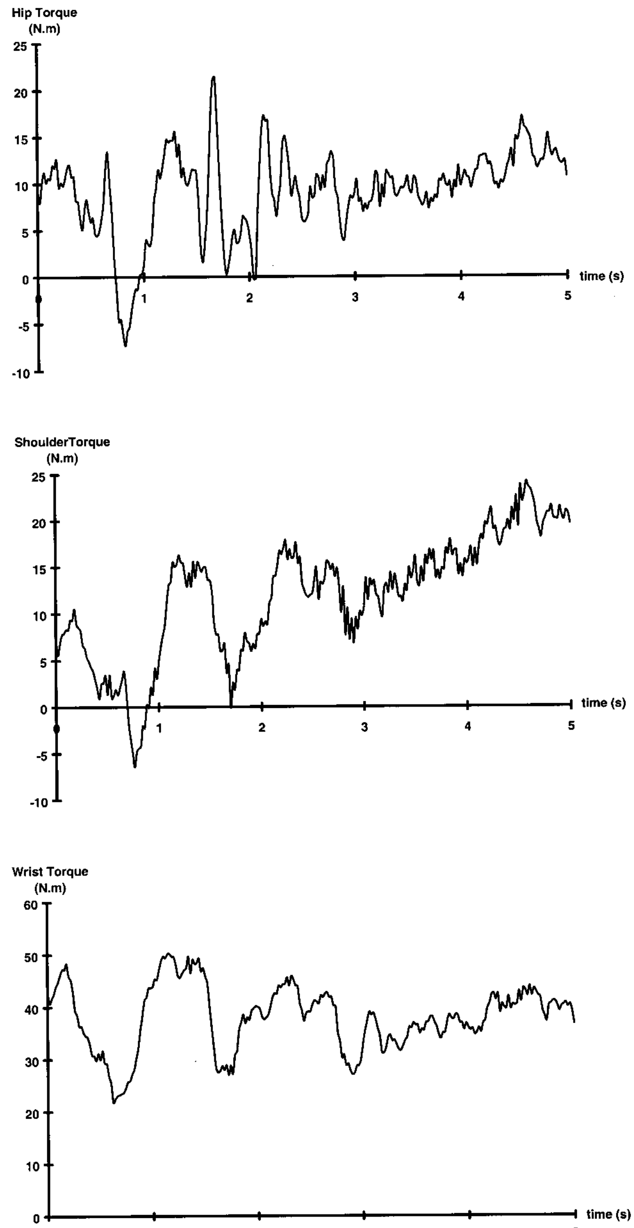

FIGURE 6—Normalized torque time histories for wrist, shoulder, and hip joints.

(Fig. 7) shows that the horizontal displacements of ankle and toe positions follow "in-phase" at almost constant differences, indicating limited joint movement. Although these translations will influence whole-body CM position, with limited specific joint movement, it is likely that the translation of the feet was initiated further down the body (e.g., hips or wrists) rather than the feet acting as independent controllers of balance. Additionally, during inverted stance, any movement about the ankle joint will only influence a small proportion of body mass, and therefore, if the measure of control is maintenance of CM within certain limits, oscillations of the feet will not exert a significant influence on the control of inverted posture.

Slobounov and Newell (12) also found that increases in body range of motion occurred in distal segments. Partial support for such findings can be observed in some trials where joint angle changes are very small for the hip and shoulder, indicating a relatively fixed trunk (Fig. 5A). However, in other trials, there was an increase in range of motion traveling up the body indicating that small movements in proximal joints (wrist, elbow) are propagated up the link system causing progressively larger degrees of movement at each joint, finally manifesting in large movements of distal segments (Fig. 5B). Therefore, rather than initiating movement specifically in distal segments, these regions of the body are influenced by movements closer to the base of support.

The discovery that for a number of trials there is a similar pattern of joint inclusion for regressions accounting for CM variance, namely wrist → shoulder → hip, would support the transference of research findings from upright standing studies if it is assumed that hand standing with its unlearned/unstable nature is similar to upright stance in the presence of perturbations. Direct comparison of the role of the ankle joint in normal stance and the wrist joint in inverted stance as initial balancing mechanisms can be assumed. Continuing comparisons can be made between the role of the hip in upright and the shoulder in inverted stance. Thereby, to correspond to the "ankle strategy" and "hip strategy" of upright stance, inverted stance has the "wrist strategy" and "shoulder strategy." Furthermore, the finding that hip torques often play a role in maintaining balance in handstand even with only self-induced body movements may provide confirmation that, due to reduced muscular strength, altered body orientation may necessitate the utilization of torques further from the support surface as predicted by Kuo and Zajac (7).

It would appear that trials assigned with higher ranks, based on measures of steadiness, possess relatively consistent characteristics with respect to joint angle ranges and joint inclusion strategies. For instance, B1 and B5, the "superior" trials, exhibit a considerably smaller degree of hip angle movement relative to shoulder angle movement when compared with other trials. Furthermore, these two trials both produce a joint order of inclusion in CM regressions of T_1, T_2, T_3. These observations may indicate that in better trials wrist torque and shoulder torque are more vital in controlling CM movement. However, as steadiness of balance deteriorates, then hip movement/hip torque becomes increasingly more significant in recovering a wayward CM position.

One point of note is that any conjectures regarding the joints responsible for maintaining handstands provided in

TABLE 2. Stepwise estimation multiple regressions.

Trial	Adjusted R^2 in Final Model	Order of Inclusion	Significance (*P*) in Final Model
B1	0.68	T_1, T_2, T_3	0.001, 0.001, 0.019
B2	0.59	T_1, T_2, T_3	0.001, 0.001, 0.001
B3	0.58	T_2, T_3, T_1	0.006, 0.001, 0.001
B4	0.60	T_1, T_3, T_2	0.001, 0.001, 0.005
B5	0.70	T_1, T_2	0.001, 0.001, (T_3 = 0.562)
B6	0.75	T_2, T_3	0.001, 0.001, (T_1 = 0.292)

T_1, wrist torque; T_2, shoulder torque; T_3, hip torque.

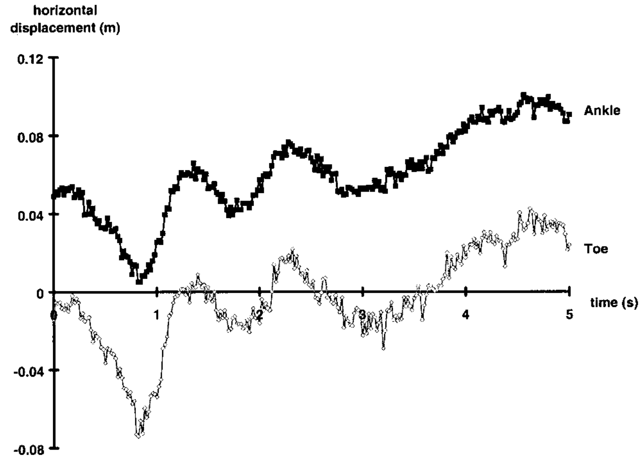

FIGURE 7—Time series of mean ankle and mean toe horizontal positions.

this study relate only to balance mechanisms acting in the anterior-posterior direction. Following the work of Winter et al. (14), with reference to alternative control strategies existing between anterior-posterior and medio-lateral balance, it is deemed inadvisable for this study to attempt to expand any results to lateral balance strategies.

A limitation of the inverse dynamics approach is that it can give no indication as to whether specific joint torques are exerted by the musculature about that particular joint or are transferred down the system by other joint torques produced elsewhere. To overcome this drawback, one possible extension of this work is the investigation of handstand balance using a forward dynamics approach where the contributions of individual joint torques could be determined.

To conclude, this study aimed to address two questions: Which joints are used in maintaining a handstand balance and are joint torques used in a strategic manner? First, in general across trials, all joint torques identified in this analysis made some contribution to maintaining anterior-posterior CM position. Second, in terms of joint order of preference, a number of trials demonstrated a similar pattern, namely wrist torque followed by shoulder torque followed by hip torque. Tentative predictions were made to suggest that superior handstand balancers (B1 and B5) may utilize sophisticated balancing strategies involving a more obviously dominant wrist torque in conjunction with influence from shoulder torques with little influence from torques about the hip joint. In contrast, the results of the stepwise regressions indicated that in handstand balances with low ranks (B4 and B6) hip torque (T_3) took on a more elevated position in the order of inclusion, moving from a position of no inclusion or third inclusion to a position of secondary inclusion. Therefore, from these initial observations there is some evidence to suggest that as balance steadiness deteriorates, hip torques become increasingly more important and the range of motion about all joints increases in an attempt to maintain an adequate body position.

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REFERENCES

1. Cohen, J. *Statistical Power Analysis for the Behavioral Sciences*, 2nd Ed. Hillsdale, NJ: Lawrence Erlbaum, 1988, pp. 407–419.
2. George, G. S. *Biomechanics of Women's Gymnastics*, Englewood Cliffs, NJ: Prentice Hall, 1980, p. 67.
3. Gurfinkel, V. S., K. E. Popov, and B. N. Smetanin. The support input as a reference for postural control. In: *Posture and Gait: Control Mechanisms*, M. Woollacott and F. Horak (Eds.). Corvallis, OR: University of Oregon Books, 1992, pp. 186–189.
4. Horak, F. B., and L. M. Nashner. Central programming of postural movements: adaptation to altered support-surface configurations. *J. Neurophysiol.* 55:1369–1381, 1986.
5. Karara, H. M. Non-metric cameras. In: *Developments in Close Range Cinematography*, Vol. 1, K. B. Atkinson (Ed.). London: Applied Science, 1980, pp. 63–80.
6. Kerwin, D. G., and A. C. Maybery. Video digitization accuracy. *J Sport Sci.* 12:171–172, 1994.
7. Kuo, A. D., and F. E. Zajac. An analysis of biomechanical constraints on the coordination of standing posture. In: *Posture and Gait: Control Mechanisms*, M. Woollacott and F. Horak (Eds.). Corvallis, OR: University of Oregon Books. 1992, pp. 344–347.
8. Kuo, A. D., and F. E. Zajac. A biomechanical analysis of muscle strength as a limiting factor in standing posture. *J. Biomech.* 26:137–150, 1993.
9. Lebiedowska, M. K., Syczewska, M., Graff, K., and M. Kalinowska. Application of biomechanical growth models of the quantitative evaluation of the motor system in children. *Disabil. Rehabil.* 18:137–142, 1996.
10. McNitt-Gray, J. L., and D. D. Anderson. Balance control strategies in upright and inverted postures. In: *Posture and Gait: Control Mechanisms*, M. Woollacott and F. Horak (Eds.). Corvallis, OR: University of Oregon Books, 1992, pp. 408–411.
11. Riccio, G. E. Information in movement variability about the qualitative dynamics of posture and orientation. In: *Variability and Motor Control*, K. M. Newell and D. M. Corcos (Eds.). Champaign, IL: Human Kinetics, 1993, pp. 317–351.
12. Slobounov, S. M., and K. M. Newell. Postural dynamics in upright and inverted stances. *J. Appl. Biomech.* 12:185–196, 1996.
13. Stephens, M. J., J. S. Frank, A. L. Burleigh, and D. A. Winter. Mechanical properties of postural strategies in controlling erect stance. In: *Posture and Gait: Control Mechanisms*, M. Woollacott and F. Horak (Eds.). Corvallis, OR: University of Oregon Books, 1992, pp.432–435.
14. Winter, D. A., F. Prince, J. S. Frank, C. Powell, and K. F. Zabjek. Unified theory regarding A/P and M/L balance in quiet stance. *J. Neurophysiol.* 75:2334–2343, 1996.
15. Wood, G. A., and L. S. Jennings. On the use of spline function for data smoothing. *J. Biomech.* 12:477–479, 1979.
16. Yeadon, M. R. The simulation of aerial movement: I. The determination of orientation angles from film data. *J. Biomech.* 23:59–66, 1990.
17. Yeadon, M. R. The simulation of aerial movement - II. A mathematical inertia model of the human body. *J. Biomech.* 23:67–74,1990.