The role of anticipatory postural adjustments during whole body forward reaching movements

THE purpose of this study was to examine the role of anticipatory postural adjustments (APAs) in the execution of forward oriented whole body reaching movements. From the standing position, eight healthy subjects were asked to reach an object placed at 45 cm from the feet, at both naturally paced and fast speeds. Electromyographic signals of six antagonistic muscles were analysed in conjunction with centre of mass (CM) displacements, centre of foot pressure displacements and resultant ground reaction forces. Results revealed that APAs created necessary angular momentum of body segments for effective task execution. These results suggest that APAs can initiate movements conducted from a fixed base of support, and in this context do not act solely to stabilize the CM. *NeuroReport* 9: 395–401 © 1998 Rapid Science Ltd.

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Introduction

The execution of voluntary movement evokes both inertial and gravitational forces which acting upon the segment chain, may disturb the position of the body centre of mass (CM). A great deal of experimental evidence, during for example arm[1,2] and leg raising,[3] forward and backward bending,[4,5] or rising onto tiptoes[6] has, through electromyographic (EMG) studies outlined postural activity preceding movement onset. Such 'anticipatory postural adjustments' (APAs) are characterized by an early inhibition of the soleus (SOL) muscle followed by an activation of the tibialis anterior (TA), producing a displacement of the centre of pressure (CP) backwards. In tasks where the overriding goal is to conserve CM projection within a limited base of support, APAs have been assumed to be programmed in order to counteract net forces arising from focal movement execution, thus stabilizing the position of the CM.

Although a task requiring the CM to leave the base of support, the study of gait initiation has led to the attribution of a different initiatory role to APAs. Here, similar patterns of anticipatory ankle muscle activity in both SOL and TA also caused CP shifts backwards and towards the stepping foot, provoking a forward fall inducing initial body velocity.[7] Thus, the true role of APAs in voluntary movement execution seems unclear. Additionally, experimental results[8] have shown that the initiation of a large repertoire of forward oriented movements (walking, rising onto tiptoes, forward trunk bending, arm raising and sit-to-stand) regardless of base of support modifications or the movement goal were accompanied by the SOL-TA inhibition/activation and CP relationship. It seems surprising that this identical EMG pattern can be attributed to both the initiation of CM displacement out of the supporting foot area during walking for example, and CM displacement minimization within the base of support during for example, arm raising.

This study aimed to re-examine the role played by APAs during the coordination between posture and movement. We have based such a re-examination upon results from a whole body reaching (WBR) task that commonly demonstrates axial synergy type opposing displacements of the trunk and knees to the hips, similar to those previously described during forward and backward trunk bending.[4,5] The majority of tasks used to study the coordination between posture and movement have involved voluntary movements being imposed upon essentially 'static' postures. During WBR tasks, where subjects reach towards an object placed on the ground in front of them, large limb displacements may indeed create conditions of disequilibrium. Thus, the central

nervous system has to solve conflicting goals of correct arm trajectory formation and equilibrium maintenance.

Here, we intend to show that when subjects conduct forward oriented WBR tasks, APAs serve a role of initiation creating angular acceleration of body segments and CM displacement within the base of support.

Materials and Methods

Eight healthy subjects (five males and three females, mean (± s.d.) age 26.5 ± 2.7 years, height 1.68 ± 0.05 m, weight 63.1 ± 7.2 kg, footlength 371.4 ± 59.3 mm) were asked to reach and lift an object to shoulder height, finishing with the arms held out horizontal in front of the body. Subjects signed consent forms and experiments were conducted in accordance with legal requirements and international norms. Two conditions of execution were imposed, naturally paced and as fast as possible. The object (a wooden dowel, mounted upon two supports weighing 1.5 kg) was placed at a distance of 45 cm from the subjects' feet in the sagittal plane. This distance corresponded to ~30% of mean subject height. Subjects began in an upright position, the hands crossed together in front of the body at hip level, and upon hearing a tone, marking a period of 500 ms after acquisition onset, reached, grasped and lifted the object to shoulder height. No specific instructions were given to subjects concerning the manner in which they were required to reach the object. Criteria for trial elimination were use of a simple 'hip' strategy (flexion of the body only at hip level, 'locking' the knee), and where the CP became situated between supporting foot area and the object at object attainment ('unbalanced trials').

Eleven reflective markers, 15 mm in diameter, were placed on the subjects' left side at the following sites: head (external canthus of the eye and the auditory meatus of the ear), trunk (the level of the inferior angle of the omoplate and the vertical of the axil), lower limb (the greater trochanter, the knee interstitial joint space, the ankle external malleolus, and the foot 5th metatarsophalangeal) and upper limb (acromial process of the shoulder, the lateral condyle of the elbow, the styloid process of the wrist and 5th metacarpophalangeal). A two-camera ELITE(system (BTS, Milan, Italy) was used to record and reconstruct successive images at a sampling rate of 100 Hz, in three dimensions. Relevant kinematic variables were low-pass filtered (digital 2nd order Butterworth filter, cut-off frequency 5 Hz) to eliminate high frequency noise.

Positions of the common body CM in the sagittal plane were calculated using a rigid seven-segment model. From horizontal and vertical displacements of various markers, the model consisted of the following appendicular and axial body segments: head–neck, upper trunk, abdomen–pelvis, thigh, shank, upper arm and forearm. Foot position was assumed to be bilaterally symmetrical and stationary. Using the model, the position of the CM of an ith segment with co-ordinates X_i, Y_i was calculated using the following formula:

$$X_i = X1_i + l_i(X2 - X1_i) \text{ and } Y_i = Y1_i + l_i(Y2_i - Y1_i)$$

where $X1_i, Y1_i, X2_i, Y2_i$ are co-ordinates of segment ends, and l_i is the ratio between the distance of the proximal marker to the segments CM and its length.

Co-ordinates X and Y of the total body CM were calculated using the formula:

$$X = \Sigma m_i X_i / \Sigma m_i \text{ and } Y = \Sigma m_i Y_i / \Sigma m_i$$

where m_i is the mass of the ith segment. Anthropometric parameters including segments masses and their CM positions were taken from the relevant literature.[9]

Global body dynamics were measured using an AMTI (Advanced Mechanical Technology Inc., Watertown, USA) force platform. The magnitude of the resultant ground reaction force vector (F_g) was determined from values of vertical (F_z) (minus the subjects weight), and horizontal (F_x) force by applying appropriate trigonometric formulae. The angle formed by the tangent of forces F_x and F_z was calculated at each 10 ms interval to provide the direction of F_g. The change in angular momentum of the whole body was taken as being equal to the external moment (M_{ext}), generated by the relationship $M_A \times F_z$, M_A (the moment arm) being the distance between horizontal projections of the CM and CP. Horizontal positions of the CP (taken from the platform) and the CM (derived from the model) were expressed within relative foot length, as a percentage of the distance between markers placed on the 5th metatarsophalangeal and the external malleolus of the subject's left foot.

Surface EMG signals were recorded from six antagonistic muscles of the lower leg (SOL, TA) upper leg (vastus lateralis-VL, biceps femoris-BF) and the trunk (rectus abdominis-RAb, erector spinae-ES). Pairs of silver chloride surface electrodes (Medi-Trace Pellet, Comepa, France) were attached longitudinally over the bulkiest part of the each muscle belly with a centre to centre electrode distance of 2.5 cm. For the SOL muscle, electrodes were attached medially at its protrusion below the gastromedialis. Signals were sampled at 500 Hz, high and low pass filtered at 7 and 20 Hz respectively, rectified, integrated at 25 Hz, and normalized as a percentage of their maximal values during reaching. For each trial

EMG burst onsets were identified in the first instance by visual inspection and quantitatively from the moment when activity exceeded mean tonic levels plus 2 standard deviations, over a 200 ms period between −500 and −300 ms before mechanical trace onset. Bursts were considered only if they exceeded this threshold for > 30 ms, following established guidelines.[10] Inhibitions or silent periods were also identified through visual inspection, latencies noted from when activity significantly reduced to below tonic levels for > 30 ms. The analysis of EMG signals was based upon their onset and inhibition latencies relative to the beginning of focal movement onset.

Intentional (focal) movement onset ($t0$) was established using various segmental curvilinear velocity profiles (derived from horizontal, vertical and translational velocity). From their bell-shaped characteristics, movement onsets were defined as the first 10 ms period where velocity profiles showed sustained deflections above zero. Despite inter-subject variations, from the four markers tested (head, hip, knee and wrist), all subject averages showed that during normally paced movements the head (auditory meatus of the ear marker) was the first to move. In fast conditions however, movements began at the knee (interstitial joint space marker). Onset latencies for these four segments are shown in Table 1. The end of reaching was defined as the first 10 ms period of wrist marker zero velocity (object attainment).

All dependent variables were checked for normality using a Kolmogorov-Smirnov test. As none of the parameters significantly deviated from a normal distribution, parametric Student t-tests were used to examine differences in dependent variables between naturally paced and fast trials.

Results

Figure 1 shows averaged WBR movements from one subject. Movements were always accompanied by hip displacements in the opposite direction to those of the arm, trunk and knee. The CM was displaced initially slightly forwards and then forwards and downwards. As expected, average movement times between $t0$ and object attainment significantly decreased from 1643.1 ± 211.1 ms (naturally paced) to 1350 ± 160.3 ms (fast; $p < 0.001$).
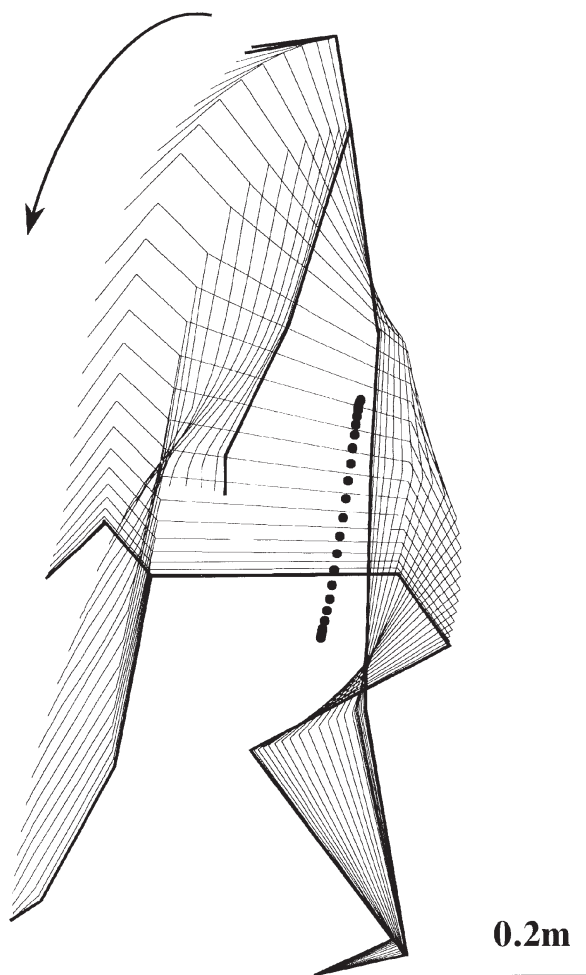

0.2m

FIG. 1. Segment displacements during whole body reaching movements (average of five trials for one subject executed at naturally paced speeds). For clarity, sticks are shown at 40 ms intervals. Segments represented are: head, head–neck, upper trunk, abdomen–pelvis, thigh, shank, foot, upper arm, forearm and hand. The centre of mass is shown as a series of filled black dots. The arrow indicates direction of reaching movements. Initial and final images are shown as slightly heavier.

EMG patterns were reproducible for all subjects and both experimental conditions. Table 2 lists average activation and inhibition latencies of all six recorded muscles in terms of $t0$. First events in both conditions were inhibitions of the tonic activity of dorsal antigravity extensor muscles, the BF and ES which were consistent in ~95% of trials. These inhibitions occurred slightly earlier during fast WBR movements. The SOL and VL showed

Table 1. All subject average (± s.d.) onset latencies (in ms) of head, hip, knee and wrist horizontal displacements during naturally paced and fast whole body reaching movements. Asterisks indicate primary segment displacement in each condition.

Condition	Head	Hip	Knee	Wrist
Naturally paced	1020.6 ± 210.1*	1132.2 ± 251.7	1085.9 ± 278.1	1130.6 ± 282.5
Fast	908.9 ± 230.1	983.9 ± 247.7	875.6 ± 254.2*	1018.9 ± 253.1

Table 2. All subject average values and s.d. of EMG activation and inhibition latencies in relation to intentional movement onset ($t0$) for all muscles in both experimental conditions. Units are in ms.

Muscle	Naturally paced	Fast
BFi	−280 ± 96.8	−301.8 ± 77.9
ESi	−270 ± 88.3	−308.3 ± 79.9
TAa	−237.5 ± 68.6	−243.8 ± 90.2
RAba	−115.7 ± 21.4	−121.2 ± 58
BFa	37.5 ± 56.5	75 ± 80.4
VLa	48.8 ± 59.6	50 ± 46.5
ESa	143.3 ± 54.2	167.1 ± 53.5
SOLa	166.2 ± 60.5	77.2 ± 64.1

Muscle name + a = activations, i = inhibitions.

gradual discharges after $t0$. Subjects demonstrated co-activations of the BF and VL muscles following the onset of intentional movement ($t0$).

Typical anticipatory postural activity in the muscles controlling the ankle (SOL and TA) is shown in relation to dynamic and kinetic events in Fig. 2. TA activation was the first detectable EMG burst, preceding prime mover (RAb) activation by around 120 ms (both conditions). There was a close temporal correlation between latencies of TA activation onset and those of backward CP displacements ($r = 0.96$, $n = 57$, $p < 0.001$). However, no evident inhibition of the SOL muscle accompanied anticipatory TA activation in the majority of trials.

The first detectable mechanical event was a backward displacement of the CP. Before this event, starting positions of the CP were significantly further forward ($p < 0.05$) within the base of support preceding fast (51.4 ± 14.9%) than naturally paced movements (41.8 ± 10.5%). During the whole of the first reaching phase horizontal CM displacements showed considerable average values of 63.8 ± 16.5 % (naturally paced, see Fig. 1) and 47 ± 13.6% (fast) of representative base of support length. They did, however, decrease significantly with movement speed ($p < 0.001$). In contrast, CP backward displacements significantly increased ($p < 0.001$), as did the magnitude of F_g ($p < 0.001$), M_A ($p < 0.001$) and the resulting M_{ext} ($p < 0.001$). Table 3 lists average values of CM, CP, M_{ext}, M_A and F_g. Thus, during fast WBR, greater M_A differences resulted primarily from larger backward CP displacements.

Figure 3 shows displacements of the CM and CP within the foot length. Also plotted is the direction of the resultant ground reaction force (F_g) from its origin (the CP), in relation to the CM. During

the phase preceding the onset of backward CP displacements, F_g was continuously aimed towards or in front of (fast conditions), CM position. Once the CP had begun its displacement, F_g passed behind the CM. Primary segment onset ($t0$) followed backward CP displacement by average values (all subjects) of 65.5 ± 17.7 ms (naturally paced) and 112.4 ± 44 ms (fast). The difference between the CM and CP (M_A) and the resultant force vector (F_g) caused a negative external moment (M_{ext}) acting in a counter-clockwise direction. As M_{ext} is known to be equal to the rate of change of whole body angular momentum,[11] it resulted here in body segment displacements, and thus the CM, forwards.

Discussion

This study examined the role of APAs preceding WBR tasks executed at two different speeds. We asked whether anticipatory muscular activity played a role of initiation, creating necessary conditions for segmental angular acceleration and CM displacements forwards. Despite obvious task differences, EMG patterns showed APAs similar to those recorded during previous studies of forward trunk bending movements.[4] Indeed, Crenna et al.[4] showed that $t0$ was preceded by clear inhibition of BF and ES and the activation of TA and RAb muscles. Our results from WBR (Fig. 3) confirmed theirs, that during fast as opposed to naturally paced movements (slow or medium paced in trunk bending), earlier TA activation and onset of mechanical trace displacements were recorded relative to $t0$. In addition to anticipatory TA activation, extensor muscles (BF, ES) showed earlier inhibition latencies during fast movements, perhaps to take advantage of gravity forces and achieve a rapid transition of posture.[12] BF and ES inhibitions also acted with earlier RAb activation to produce required forward trunk flexion. The muscle activation patterns seen after $t0$ (BF, VL, ES and SOL, although not necessarily in that order), may have been to brake forward trunk movements[4] (ES, BF + VL), or prepare the subsequent lifting movement[13](SOL).

These results from WBR would thus indicate a replication of findings from a wide range of forward oriented movements.[8] Although correlations were not made in terms of amplitude, the close temporal relationship found between latencies of TA activation and backward CP displacement onset seems to

FIG. 2. Integrated soleus (SOL) and tibialis anterior (TA) electromyographic activity (**A**) during whole body reaching movements presented with corresponding centre of mass (CM), centre of pressure (CP) displacements (**B**), and the resulting external moment (**C**). Data represent averages of all subjects and both experimental conditions. In (A) shaded areas indicate anticipatory activation of the TA muscle. Scales for each muscle are shown as a percentage of maximum burst activity. In (B) CM displacements are shown as thick black lines and CP, thin black lines. In (C) negative values represent moments applied in a counter-clockwise direction with respect to the CM. In graphs B and C, s.d. error bars are shown at 100 ms intervals. (In all Figs. vertical lines indicate focal movement onset.)
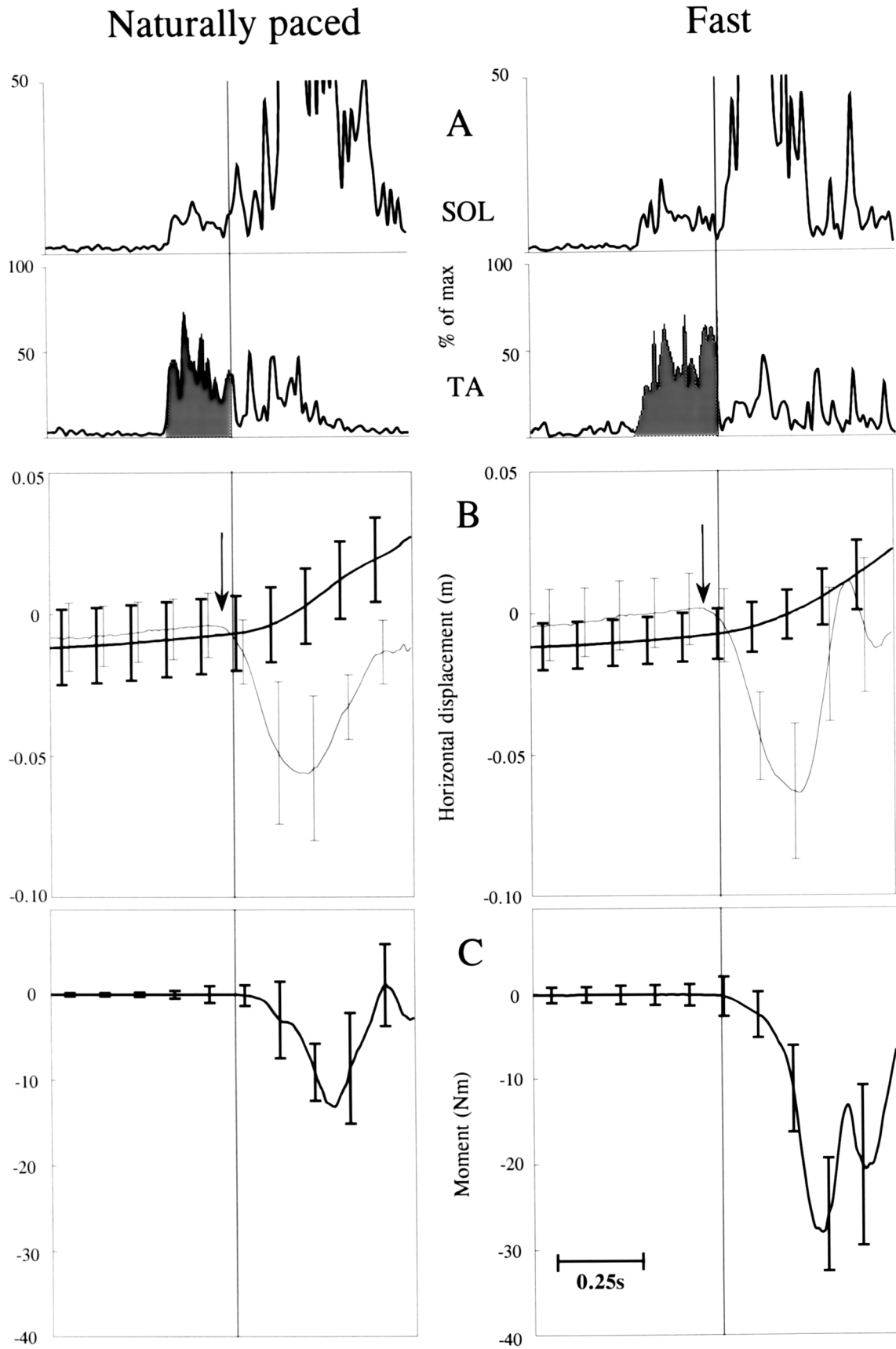
Table 3. Average values and s.d. of kinematic and dynamic measures during whole body reaching at both speeds.

	Naturally paced	Fast	p value	
			KS test	t-test
CM (% foot)	63.8 ± 16.5	47 ± 13.6	n.s.	< 0.001
CP (% foot)	39.6 ± 17.9	53.4 ± 10.8	n.s.	< 0.001
M_{ext} (Max; nm)	−11.5 ± 5.5	−30 ± 12.8	n.s.	< 0.001
M_A (Max; mm)	−72.3 ± 25.3	−101.9 ± 24.8	n.s.	< 0.001
F_g (Max; N)[a]	210.1 ± 100	433.7 ± 111.1	n.s.	< 0.001

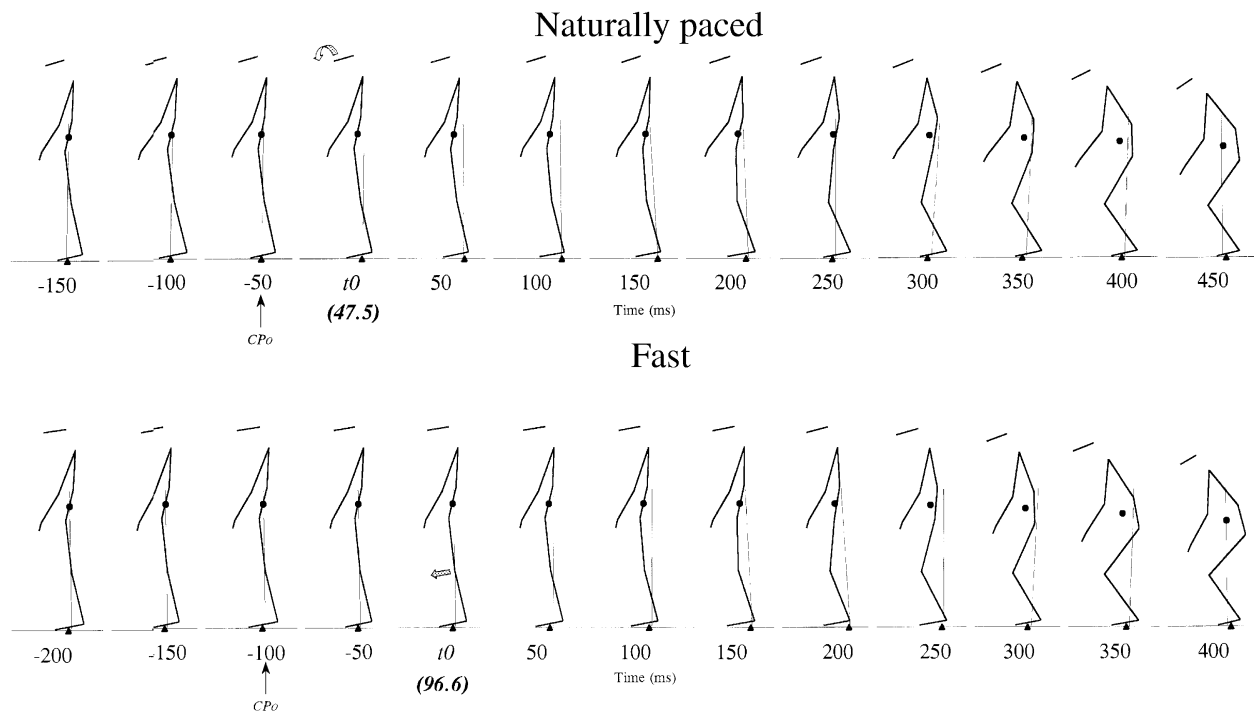[a]Minus body weight). KS = Kolmogorov-Smirnov tests of normality.

FIG. 3. Stick diagrams showing the initiation of whole body reaching movements from five trials averaged for one subject at 50 ms intervals in both experimental conditions. For clarity, a period of 600 ms with 150–200 ms before *t0* (depending on speed) is shown. Latencies between onsets of mechanical traces (backward centre of pressure displacement-CPo) and intentional movement (*t0*) for this one subject are shown in brackets. Shaded arrows indicate segment onset location (head or knee). The resultant force vector F_g (full line) is shown in relation to centre of mass position (filled black dot). The horizontal position of the centre of pressure is shown as a filled black triangle.

agree with the idea that the observed CP shift could be ascribed to anticipatory activity of the TA. However, an inhibition of the SOL muscle was not recorded, the reason for which was unclear. One possibility may be that increases in ankle stiffness, by a coupling of TA activation and tonic SOL activity, created a stable mechanical base ensuring postural stability during the anticipatory phase before forward acceleration of the CM. However, the possibility of cross-talk from the highly active TA muscle cannot be ruled out.

During the entire reaching phase, horizontal CM displacements showed amplitudes that attained values of up to 96.4% (naturally paced) and 72.7% (fast) of relative footlength. This indicates displacements significantly greater than those recorded during other whole body movements.[3–6] CM amplitudes within the BS, in the present study, suggest it difficult to accept that the primary role of the APAs was to minimize their displacements by creating inertial forces equalling those produced by the voluntary movement itself.[2] An alternative explanation concerning the role of APAs may be proposed in light of EMG, kinematic and dynamic findings during the initiation of WBR. From our results, we suggest that the pattern of APAs preceding WBR did not serve exclusively to control for final posture,[14,15] or forthcoming movement perturbations,[16] but created segmental angular acceleration and CM displacement within the BS. This hypothesis is supported by a number of the results.

First, during the execution of WBR movements, subjects encounter conflicting dynamic equilibrium

Table 4. List of biomechanical abbreviations used in the text

CM	=	centre of mass
CP	=	centre of pressure
Fx	=	Horizontal force
Fz	=	Vertical force
Fg	=	Resultant ground reaction force vector
Mext	=	External moment
MA	=	Moment arm

and performance constraints. Hence, they are faced both with minimising horizontal momentum but producing required forward angular momentum. To fulfil these criteria, an increase in the external moment is achieved by increasing the vertical moment arm (the distance between vertical projections of the CM and CP in the horizontal axis), which is under primary control from the CP. Despite their large displacements within the BS, findings that CM displacements throughout the entire reaching phase decreased with increasing speed may be explained by considering the relationship between horizontal and vertical momenta. Backward CP displacements create an increase in the vertical moment arm, fulfilling the need both for increased angular momentum (and movement speed) and the conservation of equilibrium. As ankle muscular activity has been shown to control displacements of the CP throughout a wide range of voluntary arm and trunk bending movements,[8] we suggest backward CP displacements in relation to CM position at the initiation of WBR, created the necessary external moment thus serving a focal role of initiating reaching movements. This relationship satisfactorily fulfils constraints of both equilibrium and performance.

Second, during this particular WBR task F$_g$ was not continuously aimed towards the CM. Following the onset of dynamic events, F$_g$ was oriented behind the CM, producing forward segmental angular acceleration.[17] Differences between the two reaching speeds illustrated that in order to increase angular acceleration, the moment arm was increased. This resulted in a negative external moment exerted in a counter-clockwise direction. As the external moment has been described as being equal to the rate of change of whole body angular momentum,[11] the pattern of APAs recorded here was programmed to initiate reaching movements involving significant segmental angular displacement forwards.

Conclusion

These results provide a new insight into the role of APAs in the co-ordination between posture and multijoint movement. APAs have been studied throughout a wide range of voluntary movements[8] and, except where the CM leaves the BS, during human gait initiation[7] for example, have been largely attributed to the minimization of CM disturbances, either through the production of counteracting inertial forces[1,2] or the shift to a position of equilibrium for efficient movement execution.[3,6] Our results showed that the pattern of APAs initiated CM displacements during reaching movements executed from initial standing postures. Despite this, we cannot rule out the possibility that, in view of the smaller CM displacements during fast reaching, postural components of the observed APA pattern existed that served to maintain CM projection within the BS. We would be inclined therefore to adopt the idea that APAs may be programmed both for movement performance and equilibrium control.

References

1. Belen'kii VY, Gurfinkel V and Pal'tsev YI. *Biofizika* **12**, 135–141 (1967).
2. Bouisset S and Zattara M. *Neurosci Lett* **22**, 263–270 (1981).
3. Mouchnino L, Aurenty R, Massion J and Pedotti A. *J Neurophysiol* **67**, 1587–1598 (1992).
4. Crenna P, Frigo C, Massion J and Pedotti A. *Exp Brain Res* **65**, 538–548 (1987).
5. Pedotti A, Crenna P, Deat A et al. *Exp Brain Res* **74**, 3–10 (1989).
6. Nardone A and Schieppati M. *Exp Brain Res* **69**, 469–480 (1988).
7. Breniére Y, Do MC and Sanchez J. *J Biophys Méd Nucl* **5**, 197–205 (1981).
8. Crenna P and Frigo C. *J Physiol (Lond)* **437**, 635–653 (1991).
9. Winter DA. *Biomechanics and Motor Control of Human Movement*, 3rd edn. New York: Wiley, 1990.
10. DeLuca CJ. *J Appl Biomech* **13**, 135–163 (1997).
11. Beer FP and Johnson R. *Vector Mechanics for Engineers: Statics and Dynamics*, 2nd edn. New York: McGraw-Hill, 1972.
12. Cheron G, Bengoetxea A, Pozzo T et al. *Electroencephalogr Clin Neurophysiol* **105**, 58–71 (1997).
13. Commissaris DACM and Toussaint HM. *Hum Movement Sci* **16**, 407–431 (1997).
14. Lipshitz MI, Mauritz K and Popov KE. *Fiziol Cheloveka* **7**, 411–419 (1981).
15. Do MC, Nouillot P and Bouisset S. *Neurosci Lett* **130**, 9–11 (1991).
16. Bouisset S and Zattara M. *J Biomech* **20**, 735–742 (1987).
17. Toussaint HM, Commissaris DACM, van Dieën JH et al. *J Motor Behav* **27**, 225–234 (1995).