Biomechanical approach to quantifying anticipatory postural adjustments in the elderly

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Abstract—*The paper outlines a biomechanical approach to quantifying anticipatory postural adjustments in the elderly. The measurement problems that occur in applying the biomechanical approach to elderly subjects are described and the 'signal-to-noise' properties of three candidate measures are compared, using data from volitional unilateral arm-raise tests performed on 100 elderly subjects. The results suggest that changes in vertical ground reaction force provide the greatest potential for accurate measurement of anticipatory adjustments, in comparison with changes in horizontal force or centre-of-pressure displacement. By normalising the anticipatory change in vertical ground reaction force with respect to the vertical perturbation force induced by the arm motion, a measure of relative anticipatory response is derived. The use of this measure, as well as its limitations, are demonstrated by analysing its relationship to actual falling risk, monitored prospectively in the elderly subject population. The findings showed evidence of larger relative anticipatory adjustments in the subjects who experienced recurrent falls, and it is suggested that these responses may be indicative of disordered motor programming. However, to detect these differences, it was necessary to average responses over multiple trials and to exclude trials with very small arm acceleration or very large baseline 'noise' (associated with ongoing postural sway). The need to screen and exclude data would seem to limit the practical utility of this approach in testing elderly populations.*

Keywords—*Ageing, Balance, Biomechanics, Motor control, Postural control*

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1 Introduction

THE POSTURE control system is thought to act through two types of processes: first closed-loop (or feedback) control, in which errors between desired and actual postural state are sensed and corrected, and secondly open-loop (or feedforward) control, in which corrective responses are initiated in a predictive manner without detection of errors. One particular type of open-loop control involves the anticipatory postural adjustments that precede voluntary movement and thereby act to compensate for the postural perturbation that the movement induces.

The most widely used paradigm for studying anticipatory postural control involves measurement of the adjustments that occur in standing subjects during rapid arm raises (i.e. shoulder flexions), performed either unilaterally or bilaterally. Typically, electromyographic (EMG) measurements are used to determine the relative timing of activation of the postural leg muscles and the 'focal' arm muscles involved directly in raising the arm. Although this direct approach undoubtedly provides the most clear-cut evidence of the degree to which the postural adjustments are anticipatory in nature, the need to apply EMG electrodes can make the procedure relatively awkward and time-consuming. Furthermore, for certain subject populations, such as the frail elderly, the use of electrodes may increase anxiety which, in addition to creating practical difficulties, could also conceivably affect the measured responses (e.g. by promoting co-contraction of antagonist muscles).

There have been a small number of studies that have investigated the biomechanical correlates of anticipatory postural adjustments. These studies have shown success in demonstrating anticipatory changes in the centre-of-pressure on the feet (RIACH and HAYES 1984; 1990; RIACH *et al.*, 1992) and in the vertical and horizontal forces and rotational moments generated at the feet (BOUISSET and ZATTARA, 1981; 1987). Apparently, however, the biomechanical approach has not been widely applied to elderly subject populations, in whom the advantages of the electrode-free approach might be the most important. Furthermore, none of the previous studies has provided a means of comparing, between individuals, the relative strengths of the anticipatory postural adjustments. This is of importance if the measures are to be used in the prediction of relative falling risk, which is often one of the goals of balance testing in the elderly.

The purpose of this paper is to outline a biomechanical approach to quantifying anticipatory postural adjustments in the elderly. The measurement problems that occur in

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applying this approach to elderly subjects are described and the limitations of three candidate measures are compared, using data from unilateral arm-raise tests performed on 100 elderly subjects. A measure of relative anticipatory response is proposed and its use and limitations are demonstrated by analysing its relationship to falling risk in this elderly subject population.

2 Biomechanical considerations

In raising the arm, the initial upward and forward acceleration of the arm is due to a flexion moment generated by the shoulder musculature. As shown in the free-body diagram in Fig. 1, this creates the following sagittal-plane reactions acting on the body at the shoulder joint: a downward vertical force V_s, a backward horizontal force H_s and a forward moment M_s. If the body is modelled as a simple inverted pendulum with rotation occurring at the ankles,

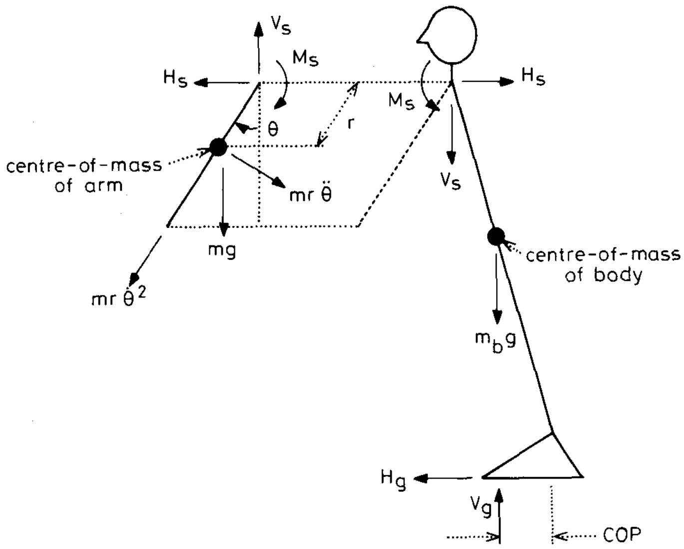

Fig. 1 *Free-body diagrams of the arm and body*
V_s *vertical force at shoulder*
H_s *horizontal force at shoulder*
M_s *net muscle moment at shoulder*
V_g *vertical ground reaction force*
H_g *horizontal ground reaction force*
θ *angle of arm with respect to vertical reference*
COP *centre-of-pressure displacement*
m *mass of arm*
m_b *mass of remainder of body*
r *distance from shoulder centre-of-rotation to centre-of-mass of arm*
g *acceleration due to gravity*

the net effect of these reactions is to rotate the body backward about the ankles (the moment about the ankle due to the backward horizontal force predominates, due to the large moment arm).

Thus, the perturbation induced by the arm acceleration acts initially to accelerate the centre-of-mass of the rest of the body downward (due to V_s) and backward (due to H_s) and to rotate the body backward (due to the net moment at the ankle). BOUISSET and ZATTARA (1987) have shown that the anticipatory postural adjustments normally act in a direction so as to oppose these perturbations. Thus, the anticipatory changes in the vertical and horizontal ground reaction forces measured at the feet are directed so as to accelerate the centre-of-mass upward (i.e. increase in V_g) and forward (i.e. increase in H_g). Similarly, RIACH and HAYES (1984) and RIACH *et al.* (1992) have shown that the anticipatory centre-of-pressure displacement is usually directed posteriorly; apparently, this is due to a reduction in tonic soleus activity, allowing body weight to create a net dorsiflexion moment at the ankle and thereby rotate the body forward. It should be pointed out that unilateral arm raises also create a rotational perturbation about a vertical axis; however, for simplicity, the focus of the present paper will be limited to disturbances occurring in the sagittal plane.

Based on the foregoing analyses, three candidate biomechanical variables were selected for investigation:

(*a*) increase in vertical ground reaction force, acting in an upward direction on the body (V_g)
(*b*) increase in horizontal ground reaction force, acting in an anterior direction on the body (H_g)
(*c*) posterior displacement of the centre-of-pressure on the feet (COP). Note that change in centre-of-pressure displacement is approximately proportional to change in net ankle moment (MAKI, 1987).

Anticipatory adjustments can be quantified by determining the change in these variables (relative to the pre-onset baseline) that has occurred at the instant when the onset of arm motion is first detected. Although later post-onset changes in the measured biomechanical variables may also be due to a preprogrammed adjustment, these post-onset changes may also result from feedback responses to the perturbation or from passive ground reaction forces induced by the perturbation.

To quantify interindividual differences in the anticipatory adjustments, the magnitude of the adjustment must be related to the magnitude of the self-induced perturbation. As is demonstrated by the experimental results described below, this turns out to be particularly important for elderly subjects, who tend to show large differences in the magnitudes of arm acceleration that they are able (or willing) to generate. Modelling the arm motion as rigid-body rotation about a fixed shoulder axis (i.e. neglecting the relatively small acceleration of the shoulder itself), the perturbing shoulder forces are a function of the tangential ($r\ddot{\theta}$) and centripetal ($r\dot{\theta}^2$) acceleration of the centre-of-mass of the arm, i.e.:

$$V_s = mg + mr\ddot{\theta} \sin \theta + mr\dot{\theta}^2 \cos \theta \tag{1}$$

$$H_s = mr\ddot{\theta} \cos \theta - mr\dot{\theta}^2 \sin \theta \tag{2}$$

where r is the distance between the centre-of-rotation of the shoulder and the centre-of-mass of the arm. This distance is estimated to be equal to 0.53 times the length of the arm, and the arm mass m is estimated to be 5 per cent of the total body mass (WINTER, 1979).

To determine the perturbation forces, the time history of the angular motion of the arm must be estimated. This can be accomplished, in a relatively inexpensive manner, by using a linear accelerometer to measure the tangential acceleration at the wrist and then numerically integrating the differential equation describing the relationship between the tangential arm acceleration a_t and angular arm position θ, approximating the arm motion as rigid-body rotation about a fixed shoulder joint, i.e.:

$$r_a \ddot{\theta} = a_t \tag{3}$$

where r_a is the radial distance between the accelerometer and the shoulder axis. The tangential acceleration a_t is determined from the measured accelerometer signal a_m as follows:

$$a_t = a_m - g \sin \theta \tag{4}$$

where the second term ($g \sin \theta$) accounts for the influence of gravity on the accelerometer readings. The experimental results discussed below indicate that it is necessary to include this term in the calculations because the accelerations achieved by elderly subjects may well be of the same order of magnitude as the acceleration due to gravity; in contrast BOUISSET and ZATTARA (1987) found the influence of gravity to be negligible relative to the accelerations achieved by their 'normal adult' subjects.

3 Experimental methods

Seventeen male and 83 female subjects between the ages of 62 and 96 were tested (mean age = 83, standard deviation = 6; see Table 1 for age distribution). These subjects were participants in a prospective study of multiple predictors for risk of falling; the arm-raise tests discussed in this paper were one component of the fall-risk assessment. The sample size was selected to meet the requirements of the risk-prediction study. Volunteers were recruited from two self-care residences (residents live independently in private or shared apartments but have access to on-site nursing care and dining and recreational facilities). Volunteers were included in the study if judged able to stand unaided for 90 s and to walk 10 m using a cane or walking frame if necessary (22 individuals normally used a cane or walking frame to move about). The clinical characteristics of the subject population are summarised in Table 1; see MAKI *et al.* (1991) for detailed postural sway data.

The subjects stood on two force plates which were used to measure the horizontal (anterior-posterior) and vertical forces acting on each foot and to determine the displacement of the centre-of-pressure (COP). Static calibration tests showed the mean errors for the vertical force, horizontal force and COP measurements to be within 0.5 per cent of the reading, 1·5 per cent of the reading and 0·03 mm, respectively; error standard deviations were within 0·3 per cent, 4 per cent and 0·4 mm, respectively. The force plate design and performance is described in more detail elsewhere (MAKI, 1987; MAKI *et al.*, 1987). A miniature uniaxial, piezoresistive accelerometer {Endevco, model 2265-20} was worn in a Velcro cuff placed at the wrist, aligned so that its active axis was orthogonal to the longitudinal axis of the arm (within the sagittal plane). This accelerometer has a range of $\pm 20g$ and a frequency response of 0–200 Hz, with a combined nonlinearity and hysteresis error less than or equal to ± 2 per cent of the reading. The force plate and accelerometer signals were amplified, low-pass filtered (second-order Butterworth; -3 dB at 10 Hz) and sampled at a rate of 50 Hz.

The subjects were instructed to stand relaxed, with feet comfortably spaced (unshod) and hands at sides. They faced a computer-controlled 'trigger light', and were instructed to raise one arm to shoulder level as quickly as possible when the light was activated, to hold it at that position, and then to lower it when the light was extinguished. They were allowed to use whichever arm they preferred. Practice trials were performed until the experimenter was satisfied that the subject understood the procedure. Six experimental trials were then performed at unpredictable intervals ranging from 2 to 3·5 s; for each

Table 1 Clinical characteristics of the subject population

Clinical characteristics	Proportion of subjects (percentage)*	
	Males	Females
Age distribution:		
62–69 years	1/17 (6)	2/83 (2)
70–79 years	4/17 (24)	22/83 (27)
80–89 years	10/17 (59)	48/83 (58)
90–96 years	2/17 (12)	11/83 (13)
Measured impairments:		
balance (maximum one-leg stance duration <5 s)	10/14 (71)	63/75 (84)
balance (performance score <20/24; see TINETTI, 1986)	4/16 (25)	39/83 (47)
visual acuity (better eye, with corrective lenses, Snellen chart score <20/40)	11/17 (65)	54/82 (66)
vibration sense (errors in sensing 256 Hz tuning fork vibration at toe, ankle and shin)	8/14 (57)	44/78 (56)
kinaethesis (errors in sensing passive toe or ankle flexion/extension)	0/16 (0)	3/77 (4)
range of motion (hip flexion <60°, knee flexion <90° knee extension <−20°, ankle flexion <10° or ankle extension <20°)	4/17 (24)	7/82 (9)
grip strength (Jamar dynamometer, <20 per cent body weight)	2/17 (12)	19/83 (23)
cognition (Mini Mental State score <21/30)	0/15 (0)	0/81 (0)
Medical history:		
vertigo, dizziness or vestibular disorders	1/17 (6)	8/79 (10)
neurological/neuromuscular disorders	9/17 (53)	23/79 (29)
cardiovascular disorders	8/17 (47)	40/79 (51)
orthopaedic disorders	9/17 (53)	55/79 (70)
metabolic disorders	6/17 (35)	17/79 (22)
psychiatric disorders (including depression)	1/17 (6)	7/79 (9)
Use of medications:		
sedatives/hypnotics	2/17 (12)	24/83 (29)
tranquillisers	0/17 (0)	0/83 (0)
antidepressants	2/17 (12)	12/83 (14)
anti-Parkinsonian agents	1/17 (6)	0/83 (0)
diuretics	4/17 (24)	29/83 (35)
other antihypertensives	1/17 (6)	7/83 (8)
other cardiac medications	8/17 (47)	33/83 (40)
antiinflammatory/analgesic agents	9/17 (53)	55/83 (66)
antiseizure medications	1/17 (6)	3/83 (4)

* variation between measures in total number of subjects is due to missing data (i.e. subjects unable or unwilling to perform test or medical history not available)

trial, the light remained on for 3 s. Throughout the session, the subjects were repeatedly exhorted to raise their arm as quickly as possible. All trials were videotaped to allow possible irregularities to be assessed.

The three candidate variables V_g, H_g and COP were compared in terms of a 'signal-to-noise' ratio (SNR). The 'noise' amplitude was defined to be the peak-to-peak range of the signal recorded over a 400 ms 'baseline' interval, starting 500 ms prior to the onset of arm movement (the 100 ms interval immediately preceding arm motion was excluded to prevent anticipatory changes from affecting the baseline estimates). The 'signal' amplitude was defined to be the peak-to-peak range of the signal recorded over a 1·0 s interval following the onset of arm movement. Onset of arm movement was defined to occur when the measured arm acceleration exceeded $0{\cdot}4\,\mathrm{m\,s^{-2}}$. Calculation of V_g SNR is illustrated, for a sample trial, in Fig. 2.

For each variable, SNR scores were calculated for all trials by all subjects. A repeated measures analysis of variance (ANOVA) was then performed (with blocking on subjects) to compare the three biomechanical variables. The Waller-Duncan test was used to perform pairwise comparisons between the means, in order to rank the variables.

As detailed below, the vertical ground reaction force signal V_g proved to have the best 'signal-to-noise' properties; therefore, this variable was selected to construct the relative response measure. The change in vertical ground reaction force ΔV_g occurring at onset of arm motion was determined, relative to the average baseline level. This anticipatory change was then normalized with respect to perturbation magnitude by dividing by the peak downward perturbation force ΔV_s induced at the shoulder by the arm acceleration, thereby yielding a relative anticipatory adjustment score, $AA_{rel} \equiv \Delta V_g/\Delta V_s$. The vertical perturbation force was estimated in the manner described in the previous section, i.e. by integrating eqns. 3 and 4 to determine the angular arm kinematics from the measured accelerometer signal and then using these kinematic data to solve eqn. 1. Hammings modified predictor corrector method and a commercially available software package {SIMNON; Lund Institute of Technology, Lund, Sweden} were used to perform the numerical integrations. Fig. 2 illustrates how the relative anticipatory response measure is determined for an individual trial.

For each subject, the AA_{rel} scores from the six repeated trials were averaged so as to yield a mean relative anticipatory adjustment score $\overline{AA}_{rel}$ that could be used to characterise the subject's performance. ANOVA on the $\overline{AA}_{rel}$ scores was then performed to compare subjects who experienced no falls during a one-year prospective monitoring period ('nonfallers') and subjects who reported experiencing one or more falls during the same period ('fallers'). The analysis was repeated comparing 'recurrent fallers' (two or more falls) with the remaining subjects (fewer than two falls). Fear of falling ('fearless' against 'fearful') was included as a factor in the ANOVAS because previous work has suggested that some balance measures may be more closely related to the fear, rather than the actual risk, of falling (MAKI *et al.*, 1991). Falls were monitored via weekly self-reports; each week, subjects who failed to report were contacted by telephone. In addition to the above analyses, a two-way ANOVA was performed to assess possible gender- or age-related differences; to perform this analysis, the subjects were divided into three age groups of approximately equal size, representing age ranges 62–79, 80–86 and 87–96.

In the ANOVAS, log or rank transformations were per-

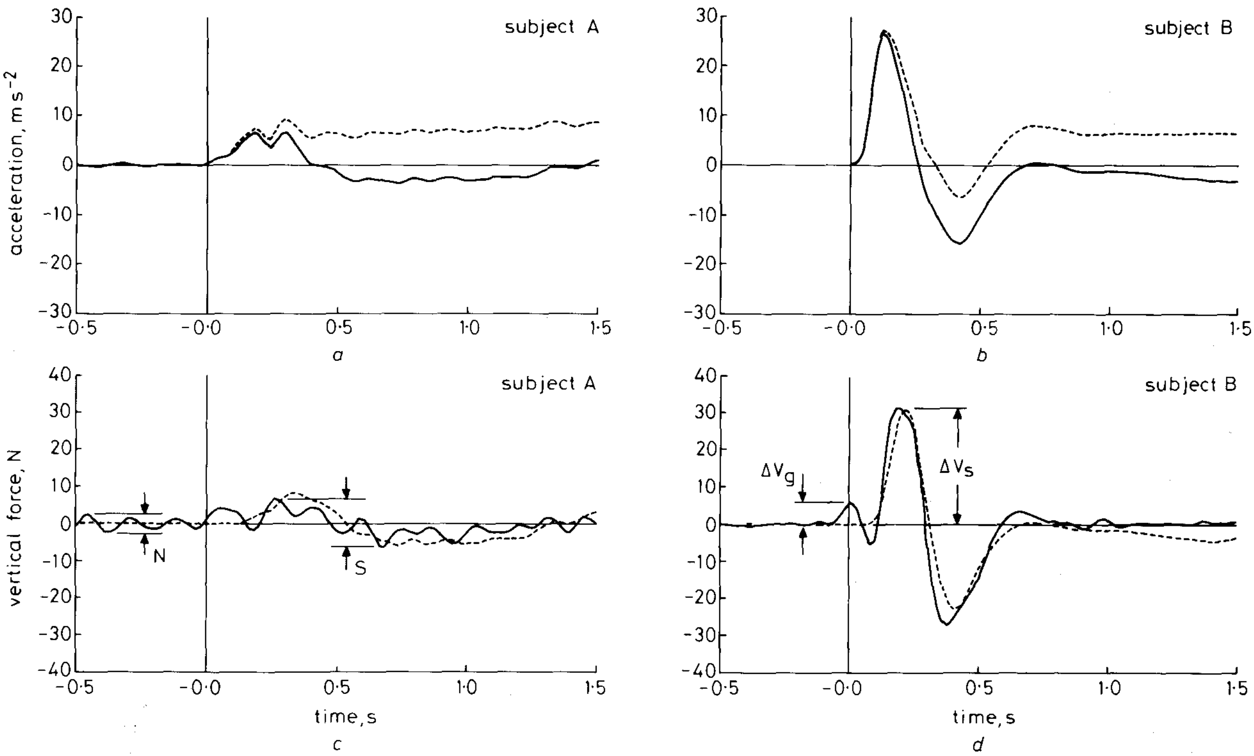

Fig. 2 *Sample data from two trials. (a) and (b) display tangential arm acceleration as measured at the wrist. The solid line is the accelerometer signal after correction for influence of gravity via eqn. 4, and the broken line is the uncorrected accelerometer signal. (c) and (d) display vertical forces. The solid line is the measured ground reaction force (minus body weight) and the broken line is the estimated perturbation force acting at the shoulder. Sign conventions are as indicated in Fig. 1, i.e. positive shoulder forces act downward on the body and positive ground reaction forces act upward on the body. (c) shows the parameters used to calculate signal-to-noise: SNR = S/N. (d) shows the parameters used to calculate the relative anticipatory adjustment score:* $AA_{rel} \equiv \Delta V_g/\Delta V_s$

formed where necessary, to 'stabilise the variance' and normalise the residuals (CONOVER and IMAN, 1981; NETER *et al.*, 1985).

4 Results

Descriptive statistics for the 'signal-to-noise ratios' (SNR) are presented in Table 2. The large standard deviations and ranges seen in these data are indicative of the large variability in this elderly subject population. Nonetheless, in spite of this variability ANOVA showed highly significant differences, on average, between the vertical force, horizontal force and COP measures ($p < 0{\cdot}0001$). Pairwise comparisons showed the vertical force V_g to provide significantly better measurement properties, on average, compared with the other variables ($p < 0{\cdot}0001$). In fact, the mean SNR for vertical force was almost twice as large, compared with the horizontal force, and four times as large, compared with the COP. On this basis, V_g was selected to construct the measure of relative response, AA_{rel}.

*Table 2 'Signal-to-noise ratios' for candidate biomechanical variables**

Variable	Mean	Standard deviation	Minimum	Maximum
Vertical ground reaction force V_g	28·1	20·4	1·4	122·0
Horizontal ground reaction force H_g	15·9	13·2	0·9	91·0
Centre-of-pressure displacement COP	8·7	7·2	0·6	58·7

* 'signal-to-noise ratio' defined as peak-to-peak range in post-onset 'signal' divided by peak-to-peak range in baseline 'noise' (see text for further details; see Fig. 2 for sample data); descriptive statistics derived from 576 trials

Two example trials, shown in Fig. 2, illustrate the range of V_g responses recorded in the elderly subject group. For subject A, the arm acceleration was very small, and the V_g data shows high levels of background (pre-onset) activity ('noise') and little discernable evidence of an anticipatory response (i.e. at time = 0). The response of subject B is more similar to that seen in 'normal adults' (BOUISSET and ZATTARA, 1987), although the acceleration and force levels are still much smaller in magnitude. Subject B also differs from the 'normal adult' response in that the initial anticipatory rise in V_g (indicating an upward acceleration of the centre-of-mass) is followed by a reduction below the baseline (indicating a deceleration of the centre-of-mass), before V_g rises again. The post-onset reversal in V_g may be a compensation for an inappropriately scaled or timed anticipatory response.

A histogram of the relative anticipatory adjustment responses AA_{rel} from the *individual trials* is shown in Fig. 3. As detailed earlier, AA_{rel} represents the magnitude of the anticipatory change in vertical ground reaction force ΔV_g expressed as a proportion of the peak perturbatory force induced at the shoulder by the arm acceleration ΔV_s. Note that 11 trials were excluded because the subjects failed to raise their arm when cued, raised their arm prematurely, raised the wrong arm or raised the arm sideways, and one trial was excluded because the subject lost her balance. Two subjects (12 trials) were excluded because of instrumentation problems.

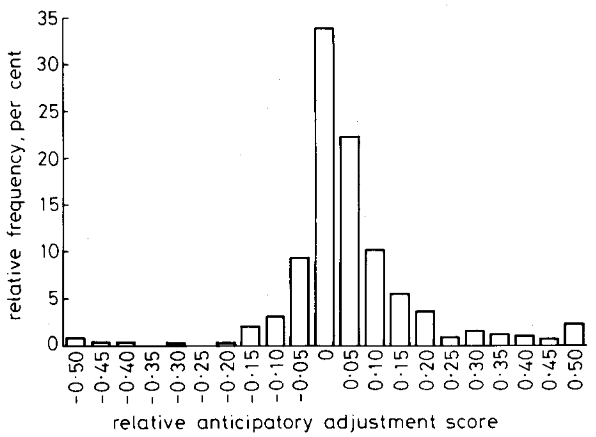

Fig. 3 *Distribution of relative anticipatory adjustment responses ($AA_{rel} = \Delta V_g/\Delta V_s$) from individual trials (total of 576 trials); extreme right hand cell include scores $>0{\cdot}525$ ($N = 7$, maximum score $= 3{\cdot}1$); extreme left hand cell includes scores $<-0{\cdot}525$ ($N = 4$, minimum score $= -3{\cdot}2$). Note: positive values indicate anticipatory responses directed so as to oppose the vertical perturbation force induced at the shoulder by the initial arm acceleration*

As illustrated by the histogram, a large number of AA_{rel} responses were very close to zero, suggesting an absence of functional anticipatory responses prior to onset of arm motion. In addition, a large population of responses were actually negative, i.e. in the direction opposite to that seen in 'normal adults' (BOUISSET and ZATTARA, 1987), indicating an anticipatory force that acts in the same direction as the perturbation rather than acting to oppose the perturbation. It should be emphasised, however, that the AA_{rel} data shown in the histogram are from individual trials, without averaging, and hence are susceptible to contamination by baseline 'noise'. Thus, although the negative responses could reflect disordered motor control, they could also be due to measurement artefact.

To minimise the influence of measurement artefacts, each subject's performance was characterised by a mean relative anticipatory adjustment score $\overline{AA}_{rel}$ which was derived by averaging the AA_{rel} scores from the individual trials. Descriptive statistics for the fall-risk/fear ANOVA on the $\overline{AA}_{rel}$ scores are presented in Table 3. The trials listed above were excluded from the ANOVA. Also excluded were four subjects who failed to complete the one-year fall monitoring period. The ANOVA of the $\overline{AA}_{rel}$ data failed to show any significant differences between the 'fallers' and 'non-fallers', or between the 'fearful' and 'fearless' subjects ($p > 0{\cdot}43$). The same was true for the analysis of the 'recurrent fallers' against the remaining subjects ($p > 0{\cdot}16$). ANOVA with respect to age and gender showed no evidence of significant age- or gender-related differences in the $\overline{AA}_{rel}$ scores ($p > 0{\cdot}30$).

Inspection of the data revealed that about 9 per cent of the trials ($N = 52$) were characterised by very small arm accelerations and almost negligible perturbation forces (i.e. less than 5 N). In approximately 8 per cent of the trials ($N = 44$), the measured anticipatory change in V_g was judged to be simply a continuation of large pre-onset background activity. After excluding these trials, as well as excluding any subjects who had fewer than three non-excluded trials ($N = 10$), the mean scores $\overline{AA}_{rel}$ were recalculated and the ANOVAs on these scores were repeated. To prevent biased decisions, the data screening was performed

Table 3 Descriptive statistics for mean relative anticipatory adjustment score $\overline{AA}_{rel}$

	Unscreened data			Screened data†		
	N	Mean	Standard deviation	N	Mean	Standard deviation
Analysis 1:						
non-fallers (0 falls)	37	0·0549	0·110	34	0·0462	0·0621
fallers (≥1 fall)	57	0·0598	0·0869	50	0·0556	0·0789
Analysis 2:						
non-recurrent fallers (0–1 fall)	72	0·0511	0·0886	64	0·0406*	0·0551
recurrent fallers (≥2 falls)	22	0·0799	0·117	20	0·0876*	0·105

† screened data: excluded trials with small perturbation force (<5 N) and trials where measured anticipatory change was judged to be continuation of background activity; also excluded subjects with fewer than three non-excluded trials

* difference between non-recurrent fallers and recurrent fallers significant, $p = 0{\cdot}02$ (rank transformed data); $p > 0{\cdot}16$ in all other comparisons; based on two-way ANOVA; factors were fall-related status and fear of falling

while blinded to the identity and status of the subjects. After making the exclusions, the results for the 'faller'/'non-faller' analysis still failed to show any strong evidence of fall- or fear-related differences; however, the 'recurrent fallers' now showed some evidence of significantly larger $\overline{AA}_{rel}$ scores, on average, compared with other subjects ($p = 0{\cdot}02$). ANOVA continued to show no evidence of age- or gender-related differences in the $\overline{AA}_{rel}$ scores ($p > 0{\cdot}28$).

5 Discussion

It is hypothesised that one purpose of the anticipatory postural adjustments is to minimise changes in the displacement of the head and trunk during the course of the voluntary arm movement. Initially, attempts were made to quantify this directly, by double-integrating the difference between the estimated shoulder force and the measured ground reaction force to determine the body centre-of-mass displacement. These attempts were unsuccessful, as the errors in the numerical integration procedures seemed to propagate and accumulate, yielding unrealistic predictions. Any future work along these lines would probably benefit by using an optoelectronic system to measure the head and trunk displacement directly.

Previous analyses of responses to externally applied postural perturbations have suggested that displacement of the centre-of-pressure on the foot, relative to the available base-of-support, may provide a measure of relative postural stability, because a more complex control strategy must be executed whenever the centre-of-pressure reaches the limits of the base-of-support (MAKI *et al.*, 1987; MAKI and FERNIE, 1988). It seems unlikely, however, that this approach will be successful in analysing the arm raise data, at least for elderly populations. The self-induced perturbations are simply too small to seriously threaten stability, hence 'sloppy' control strategies can be adopted. Furthermore, the small changes in centre-of-pressure location that result are masked, to a large extent, by the ongoing postural sway.

The 'signal-to-noise' analysis suggested that changes in vertical force would provide the most accurate measure of postural adjustments because the arm movements tended to generate much larger changes in this variable relative to the background 'noise' associated with the ongoing postural sway. One would expect greater sway-related 'noise' in the COP and horizontal force measurements, as COP displacement reflects the ankle moments that are generated during postural sway (Maki, 1987) whereas the horizontal forces reflect the fore-aft accelerations of the centre-of-mass that occur as the body sways back and forth. Conversely, the vertical acceleration of the centre-of-mass tends to be very small during quiet unperturbed stance, and hence sway-related fluctuations in the vertical ground reaction force are also small. It is interesting to note that the selection of a vertical force measure is also supported by the findings of BOUISSET and ZATTARA (1981; 1987), who reported that the anticipatory changes in vertical force tended to be larger and occurred earlier, in comparison to the horizontal force.

Note that the measured arm-raise response 'signal' was actually a combination of the true response signal plus background 'noise'. To prevent serious errors in the 'signal-to-noise' estimates, it was necessary to define parameters so as to minimise the 'noise' contribution to the measured 'signal'. Thus, because of the small magnitude of the anticipatory response, the 'signal' was defined in terms of the peak post-onset response, rather than the anticipatory response. Nonetheless, it seems reasonable to assume that, in any given subject, any anticipatory response will be approximately proportional to the magnitude of the perturbation, as reflected (approximately) in the peak post-onset response; therefore, the variable with the best post-onset 'signal-to-noise ratio' is likely to also provide the most accurate measurement of the anticipatory response.

Despite all the efforts directed at selecting the biomechanical variable with the best measurement properties, attempts to use this measure in elderly subjects were often plagued by a number of difficulties, in particular

(*a*) small levels of arm acceleration
(*b*) considerable intersubject variation in arm acceleration levels
(*c*) small or absent anticipatory responses
(*d*) large background fluctuations in the response variable.

The normalisation procedure (i.e. scaling each response with respect to perturbation magnitude) was designed to account for subject-to-subject and trial-to-trial variation in arm acceleration; however, simple magnitude scaling may fail to account for temporal factors, i.e. the timing and time history of the response and perturbation. As discussed earlier, attempts to account for these factors, i.e. by double-integrating the force time histories to determine centre-of-mass displacement, were unsuccessful because of the errors in the numerical and modelling procedures. Another limitation relates to the use of a linear scaling factor, which in fact may not be entirely appropriate. Linear scaling will fail to yield meaningful results, for example, if the arm acceleration and resulting perturbation are so small that no anticipatory responses are evoked or if the evoked anticipatory responses are too small to be measured.

The arm accelerations that the elderly subjects were able, or willing, to generate were in fact much smaller than those seen in healthy younger adults, with an average peak tangential arm acceleration of only $13{\cdot}6\,\mathrm{m\,s^{-2}}$ ($\mathrm{SD} = 5{\cdot}9\,\mathrm{m\,s^{-2}}$), compared with values of the order of $80\,\mathrm{m\,s^{-2}}$ seen in the 'normal adult' data presented by BOUISSET and ZATTARA (1987). The smaller arm accelerations in the elderly could be due to weakness or fatigue. Some of the subjects may have limited their arm acceleration to minimise pain, e.g. in an arthritic shoulder joint. Although concerted efforts were directed at exhorting the subjects to raise their arms as quickly as possible, lack of motivation may have also been a factor, particularly in view of the prevalance of depression in this type of population. Another possibility is that the elderly subjects were afraid of losing their balance. If this were the case, however, one might expect the subjects who expressed a fear of falling to generate smaller accelerations and perturbation forces; ANOVA failed to provide any strong statistical evidence to support this ($p > 0{\cdot}13$).

The anticipatory adjustments of some elderly subjects may be diminished in magnitude, or even reversed in direction, simply because of age-related deterioration in the anticipatory posture control mechanisms (INGLIN and WOOLLACOTT, 1988). The combination of smaller perturbations plus, possibly, diminished or reversed responses could explain the very small anticipatory changes seen on average in the biomechanical measures. For 'normal' (and, presumably, much younger) adults, BOUISSET and ZATTARA (1987) reported a mean value of 0·21 ($\mathrm{SD} = 0{\cdot}043$) for the peak anticipatory change in vertical ground reaction force relative to the peak post-onset change; in contrast, the mean value of this parameter for the elderly subjects tested here was only 0·034 ($\mathrm{SD} = 0{\cdot}053$).

Larger levels of background 'noise' in the elderly subjects compound the problem of measuring smaller anticipatory adjustments. These larger background fluctuations are most likely a result of the increased postural sway that occurs in ageing (MAKI *et al.*, 1990) although, in one subject, the background fluctuations appeared to be related to a tremor. The background 'noise' adds a random error to the response when it is measured at a given instant (i.e. at the onset of arm motion), but this effect was minimised by averaging the responses measured over multiple trials. On the other hand, methods which search for the minimum or maximum response (during a single trial or over an ensemble of trials) are susceptible to bias errors, as subjects who exhibit high levels of background 'noise' are more likely to record a more extreme minimum or maximum response.

In addition to the 'signal-to-noise' problems, other errors were associated with the quantification of the perturbation, which required numerical integration of the linear accelerometer signal. As a result of the double integration, even small errors in the accelerometer calibration or alignment can lead to relatively large cumulative errors in the kinematic parameters (for example, see the growing drift in the corrected tangential acceleration signal in Fig. 2*b*). In estimating the perturbation force, further errors are introduced as a result of modelling approximations (i.e. rigid-body fixed-axis rotation), as well as inaccuracies in the estimated anthropometric parameters. In future, some of these errors might be reduced by using an angular accelerometer to measure the angular arm motion directly. Although optoelectronic motion analysis systems could also be used, accelerometers have advantages in terms of lower cost and ease of use and, for these reasons, might be preferred in clinical applications.

To examine possible effects of the errors in estimating the shoulder perturbation force on the experimental findings, alternative relative anticipatory adjustment scores were calculated by simply dividing the anticipatory change by the peak post-onset change in the measured vertical ground reaction force. The peak post-onset ground reaction force provides an alternative indicator of the perturbation magnitude; however, this parameter has the disadvantage that it is not a measure of the perturbation alone, but also reflects the postural adjustments that occurred. Nonetheless, the two relative response parameters were in fact fairly highly correlated ($r = 0{\cdot}68$ for the unscreened data; $r = 0{\cdot}93$ after eliminating the trials with very small perturbations and/or large baseline 'noise') and results of the fall-risk analysis were essentially unchanged in using the alternative parameter. Thus, it would seem that any errors that may have occurred in estimating the shoulder perturbation force did not have a major impact on the experimental results.

In general, few of the elderly subjects showed evidence of the strong, repeatable anticipatory adjustments seen in younger adults. Although this may be due in part to the measurement difficulties described above, it may also be indicative of true deterioration in postural control. Even relatively young and healthy community-dwelling elderly have shown evidence of age-related deterioration in the timing and patterning of anticipatory muscle responses (INGLIN and WOOLLACOTT, 1988), and the deterioration is likely to be much greater in the older and more frail subjects tested here. These subjects may simply fail to compensate for self-induced perturbations in an anticipatory manner. Alternatively, it may be that they compensate for the deterioration in anticipatory responses by adopting a different strategy, e.g. by generalised stiffening of the skeletal linkage through co-contraction. EMG studies are needed to examine this hypothesis.

In spite of the measurement difficulties, the fall-risk analysis did produce one interesting finding: namely, that the subjects who were at risk of experiencing recurrent falls tended to exhibit larger anticipatory adjustments, relative to the magnitude of perturbation that their arm motion induced. Although such a finding could conceivably result from 'noise' or perturbation-scaling artefacts, this would seem to be unlikely, as *post hoc* analyses failed to provide any strong evidence that the 'recurrent fallers' differed from other subjects in terms of pre-onset 'noise', 'signal-to-noise ratio' or perturbation magnitude ($p > 0{\cdot}13$).

Whereas larger relative anticipatory adjustment scores would normally be expected to reflect enhanced postural performance, it may be that the anticipatory adjustments of the 'recurrent fallers' were inappropriate for the perturbations that were induced by their arm movements. This hypothesis is supported by the observation that the larger anticipatory increases in vertical force (above baseline) were often followed by a decrease in force (below baseline); the decrease in force may represent a compensation for an initial anticipatory response that is inappropriately scaled or timed with respect to the upcoming postural perturbation. Misprogrammed anticipatory adjustments could represent a fall risk factor in themselves, or could be markers for risk factors related to other postural or motor control deficits.

In conclusion, however, it must be emphasised that fall-related differences were revealed only when results were averaged over multiple trials and when careful attention was paid to eliminating trials with very small arm acceleration or very large baseline postural sway. Use of more sophisticated motion measurement systems might help to reduce some of the measurement errors inherent in the current methodology; however, high levels of baseline sway and large intersubject variability in arm acceleration remain as more fundamental problems. Ultimately, the need for careful data screening would seem to limit the practical utility of the outlined biomechanical approach for this type of subject population.

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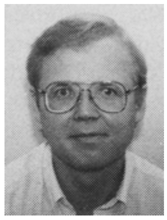
Authors' biographies



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