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BIOMECHANICS OF REACTIONS TO IMPENDING FALLS

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Abstract—Responses of 11 young adult males, initially standing, to support surface forward accelerations of 0.18 g were investigated. In response to the impending falls this stimulus initiated, body segment motions and myoelectric activities in six muscles were measured. These measurements were then input to 9 or 12 segment whole body biomechanical models and the reaction joint torques needed to produce the motions were calculated.

Mean relative joint rotations were as large as 92.8° and calculated relative joint angular accelerations as large as 29.7 rad s^{-2}. Mean myoelectric signal latencies in the six muscles monitored ranged from 135 ms at the ankles to 176 ms at the shoulders with intermediate values at intermediate joints. Mean values of calculated maximum joint torques ranged to 70 Nm at the ankles, 82 Nm at the knees, 73 Nm at the hips, and 19 Nm at the shoulders.

INTRODUCTION

Unintentional falls and jumps caused over 12,000 deaths in the United States in 1982 (National Research Council, 1985). Falls are the largest single cause of accidental death in the elderly and account for over half of all these deaths. The death rate from unintentional injury per 100,000 persons is 57 for all age groups combined, but rises to 93 for persons of ages 65–74, and 625 for persons of ages 85 or older (Kalchthaler *et al.*, 1978). Prudham and Evans (1981) estimated in their study of 2793 persons aged 65 and over an annual prevalence rate of falls, in terms of persons, of 28%. Gryfe *et al.* (1977) found an annual fall rate of 668 incidents per 1000 persons in their five-year prospective study of an ambulatory institutionalized population over 65 yr of age.

In quantitative terms, why do the elderly fall more often than the young? Does this result from declines in muscular strengths and joint ranges of motion, increasing latencies in motor response times, the development of inappropriate control of motor control programs, or combinations of these or other mechanisms?

Whatever the underlying neurologic mechanisms, problems of responses to impending falls ultimately are biomechanical problems. The masses and inertias of body segments must be supported and moved by the skeletal system through its muscular actions. Changes in cognition and central nervous system processing; changes in visual, vestibular and proprioceptive sensing abilities; the effects of physical inactivity and disuse; the effects of neurologic and orthopedic pathologies; the effects of alcohol and medications; and the effects of motivation, caution and fear all must ultimately express themselves as changes in biomechanical response. Despite this, relatively little attention has been paid to the biomechanics of impending fall responses. There are at least two reasons to study fall response biomechanics. Such studies will provide guidelines for the development of intervention therapies and valuable insights into the functioning of the musculoskeletal motor control system.

Studies to date of whole body responses to unexpected disturbances have been limited in scope. A number of laboratory studies of fall responses are reported in the literature (e.g. Melville-Jones and Watt, 1971; Greenwood and Hopkins, 1976; Do *et al.*, 1982), but these provide only isolated measures of response biomechanics. The first two of these papers report only myoelectric responses in leg muscles, and the last only floor reaction forces. Studies of reactions to perturbations of standing posture and gait (e.g. Nashner, 1976; Berthoz *et al.*, 1979; Nashner, 1982; Berger *et al.*, 1984) also provide only a few measures of response biomechanics. Biomechanical analyses of body sway while standing (e.g. Hemami and Golliday, 1977; Koozekanani *et al.*, 1980) have used models incorporating only a few body segments. Passerello and Huston (1971) constructed a ten segment whole body response model, but used it to analyze only free falls in space. A more comprehensive review of the relevant literature is provided by Romick-Allen (1986).

The present study of impending fall biomechanics in young adult males was a first attempt at a comprehensive investigation of whole body responses to unexpected disturbances. Among other things, it was pursued to develop techniques and explore requirements for an extensive study of fall response biomechanics in the elderly. Kinematic and myoelectric impending fall responses of 11 standing subjects to support surface anterior accelerations of 0.18 g were experimentally observed. These observations were input to multiple-segment biomechanical models and response kinetics analyzed.

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EXPERIMENTAL METHODS

The experimental and analysis methods used will be described only in brief. Full details are available in dissertation form (Romick-Allen, 1986).

Fall responses of 11 male subjects of ages 20–35 yrs (mean age 25.4 yrs) were observed. All subjects were in good health and reported no notable history of falling.

Fall responses were elicited by having the subjects stand on a platform whose horizontal motion was controlled. Subject safety was ensured by the wearing of a chest harness elastically suspended from overhead. After a random time delay of up to 10 s, the platform was given a nearly constant acceleration, either anteriorly or posteriorly, but with direction initially unknown to the subject. The platform then moved with a nearly constant velocity, after which it was brought to a stop.

Ten–15 tests per subject were run. These involved different magnitudes and directions of platform acceleration, presented to the subjects in randomized sequences to minimize learning effects. These sequences included two tests per subject in which acceleration direction was anterior and acceleration magnitude was 0.18 g. Only those 22 responses were analyzed. This choice was somewhat arbitrary. An acceleration of 0.18 g was just below the maximum acceleration the platform was capable of, but provoked at least moderate responses from all subjects. In those 22 tests, the platform was accelerated for approximately 160 ms, and then moved at a velocity of 29.5 cm s^{-1} for approximately 300 ms. Only the first 500 ms of the responses were analyzed to avoid the complexities introduced by platform deceleration.

Subject body segment motions were monitored by videotaping, using a single camera viewing the subject's right side at a rate of 33 frames s^{-1}. Markers were placed on the right side over the ankle joint; just below and just above the knee; posteriorly and anteriorly on the pelvis, the abdomen, and the thorax; on the superior and posterior aspects of the shoulder; and over the elbow and the wrist (Fig. 1). The locations of each of these markers in each of the approximately 16 frames of available video data were digitized for later analysis.

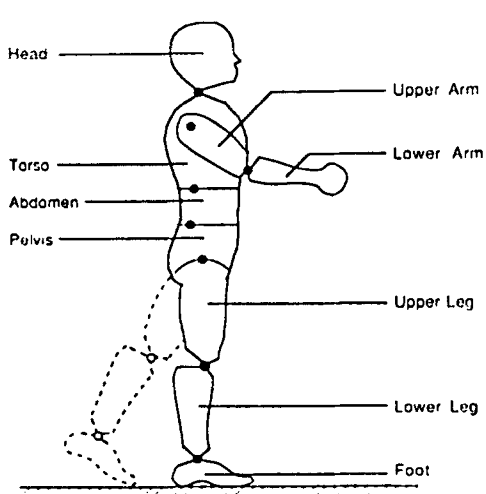


Fig. 2. Schematic diagram of the 9 and 12 segment biomechanical models used to calculate response kinetics.

Myoelectric activities were measured by bipolar surface electrodes over six right-side muscles: tibialis anterior, vastus lateralis, erector spinae, rectus abdominus, latissimus dorsi and pectoralis. Signals were amplified, filtered and recorded on chart recorders. Overall amplifier gain was 8125 with common mode rejection. Band filters passed 30 to 250 Hz, and low pass filters had a 20 ms time constant. Chart recorder response at full scale was 30 Hz. In the analyses, the only myoelectric data used were signal latencies and times over which significant co-contractions of muscle pairs occurred.

BIOMECHANICAL MODEL ANALYSIS

Fall response kinematics (motions) were experimentally measured as just described. Fall response kinetics (principally, net torques vs time at key body joints) were calculated. This was done using 9 and 12 link biomechanical models. The body segment links modelled (Fig. 2) consisted of the feet, lower legs, upper legs, pelvis, abdomen, upper torso, head and neck, upper arms, and lower arms and hands. These links were joined by single revolutes, so that movements were assumed to occur only in the sagittal plane. For analysis of responses in which one leg was stepped posteriorly, left and right foot, lower leg and upper leg segments were modeled (12 link, 11 joint model). Otherwise, sagittal symmetry was assumed (9 link, 8 joint model). Link lengths, masses, center of mass locations, and moments of inertia were assigned to the model links using literature data (McConville *et al.*, 1980; Baughman, 1983).

The observed kinematic and latency data were put into these models. If a leg was stepped, stepped kinematics were estimated from pre-and post-step leg configurations. The limited framing rate of the motion measurement system did not permit joint relative angular accelerations to be calculated by double differentiation with sufficient accuracy. Instead, joint relative angular accelerations were estimated by fitting assumed initial step followed by linear change relative acceleration waveforms (Fig. 3) to the observed body segment relative rotations by least-squares matching. Non-negligible joint relative angular accelerations were assumed to begin at the observed muscle latency times. While the joint angular acceleration waveforms used are physiologically unrealistic in that they incorporated this initial acceleration step, their convenience for data fitting overrode the small differences in calculated acceleration levels that use of more realistic waveforms would have led to.

The motions of each model segment were governed by the equations of plane rigid body dynamics. Given

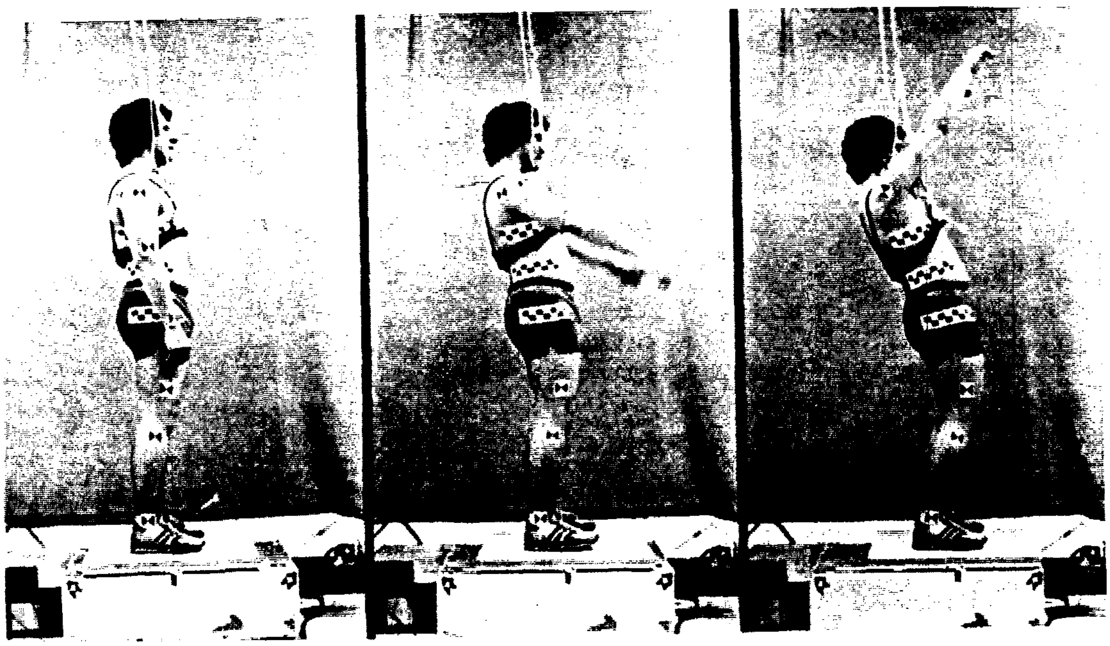


Fig. 1. Sequential configurations of a subject responding to a 0.18 g anterior acceleration of the support surface.

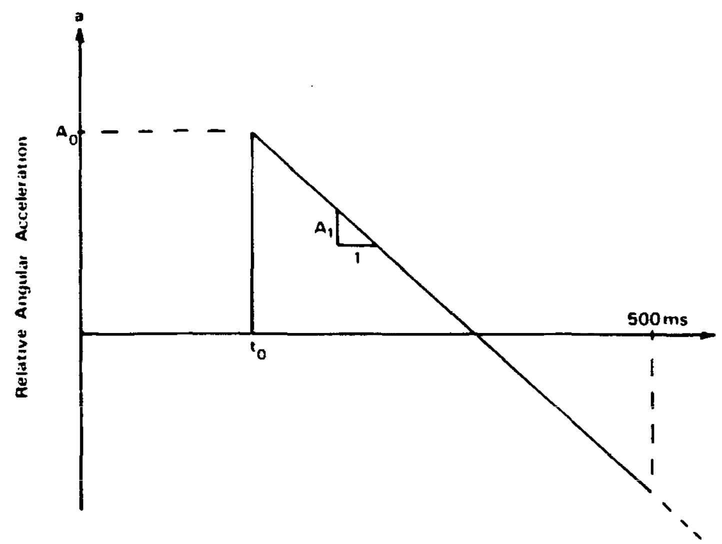


Fig. 3. Assumed form of body segment angular accelerations. t_0 = myoelectric latency time, A_0 = initial angular acceleration, A_1 = angular acceleration linear change rate.

the assemblage of the links and joints, it was convenient to express these equations of motion in Lagrange's form of d'Alembert's Principle (Huston *et al.*, 1978), so they could be automatically generated and integrated. Boundary conditions at the foot-platform interface were checked at every integration step to see if slipping or lift off had occurred. However, neither contingency arose. A standard integration algorithm was used (ISML, 1984). At each integration step, body segment accelerations, velocities and displacements and net joint forces and torques were calculated. Minimum and maximum joint torques over the observed 500 ms were noted, along with the locations of the body total mass center.

Sensitivity analyses

Limitations in the experimental capabilities available for this study prevented making direct tests of the validity of the model simulation results. As a substitute for such tests, sensitivity analyses were made. Changes in ankle and knee joint torques and in the horizontal locations of the total body mass center that resulted from 25% changes from observed or derived kinematic parameter values (latency times, initial relative angular accelerations, and relative angular acceleration change rates) were calculated using the biomechanical model.

RESULTS

Kinematic and myoelectric observations

Sixteen of the 22 observed responses to the platform anterior acceleration could be classified into one of four types. They were denoted; (1) whole body movement (WBM) response, in which substantial relative motions of all body segments occurred, including torso extension and arm swings; (2) upper body rigid (UBR) response in which the torso and other upper body segments were rotated almost as a rigid unit about the hips; (3) whole body rigid (WBR) response in which all body segment relative motions were small and (4) stepping (STEP) responses in which one leg was lifted off the support surface, swung posteriorly and then replaced on the surface. The numbers of tests in which these distinct responses were observed were 8, 4, 2 and 2 respectively. The motions in the other six responses lay intermediate to WBM and UBR responses, and so were not included when means for the different response groups were calculated.

The mean times of onset of marked myoelectric activity in the muscles monitored, over all 22 responses analyzed, ranged from 135 to 176 ms, with these mean latencies longer in the muscles crossing each more superior joint (Table 1). The trunk muscle and shoulder muscle agonist–antagonist pairs co-contracted, typically for more than 100 ms.

Body segment initial mean relative angular accelerations, in terms of the acceleration waveform fitted parameters, were as large as 24.4 rad s^{-2} (Fig. 4). Acceleration change rates varied widely (Fig. 5). The largest calculated mean relative angular acceleration

Table 1. Myoelectric signal latencies (ms). Means and S.D. over all 22 responses of the time from platform acceleration start to onset of significant myoelectric activity in the muscle (s) crossing the joints

Ankle	135	(15)
Knee	147	(18)
Torso	156	(31)
Shoulder	176	(35)

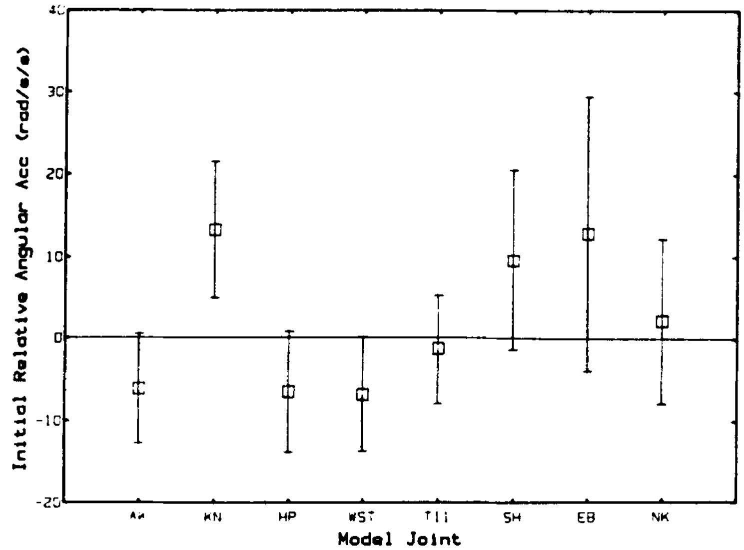


Fig. 4. Initial relative angular accelerations (rad s^{-2}) across the ankle (AK), knee (KN), hip (HP), L3 level (WST), T11 level, shoulders (SH), elbow (EB) and neck (NK). These were calculated by fitting observed relative angular displacement data with assumed kinematic wave-forms. Means and one S.D. over 22 responses are shown.

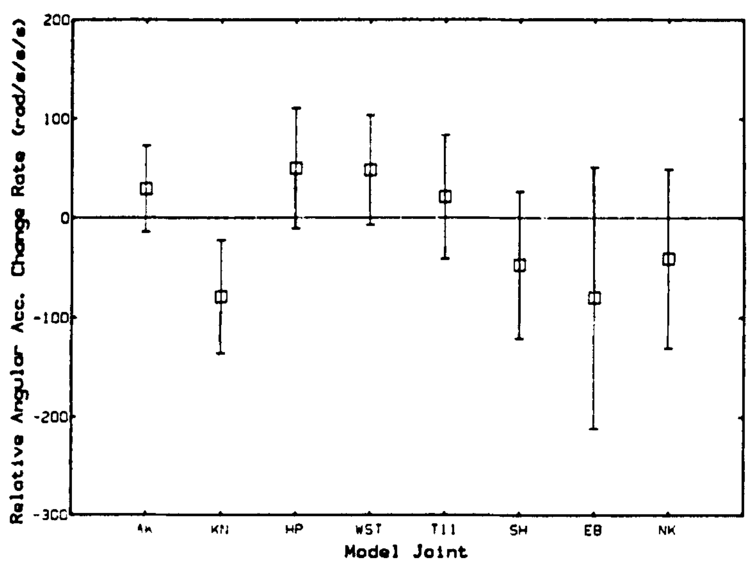


Fig. 5. Rates of change of relative angular accelerations (rad s^{-2}) across body joints. These were calculated by fitting observed relative angular displacement data with assumed kinematic waveforms. Means and one S.D. over 22 responses are shown.

and displacement were 29.7 rad s^{-2} and 92.8° respectively, and both occurred at the elbow.

Average body segment relative rotations at 500 ms after platform acceleration onset, for each type of response (Fig. 6), show that at this point, all four responses were characterized by ankle dorsiflexion; knee, hip and waist flexion and torso extension. Arm and head rotation tended to be in flexion, but this did not consistently occur.

In all of the responses, relative to the heels, the body total mass center moved first posteriorly and inferiorly and then anteriorly and superiorly (Fig. 7). It did not move beyond the heels and by 500 ms, it had returned approximately to its original horizontal location.

Computed kinetics

The means of the peak model-computed joint torques ranged to 70 Nm at the ankles, 82 Nm at the knees, 73 Nm at the hips and 19 Nm at the shoulders (Fig. 8). The initial torque discontinuities in the Fig. 8 curves are artefacts that arose from use of the assumed acceleration waveforms. The stepping response, com-

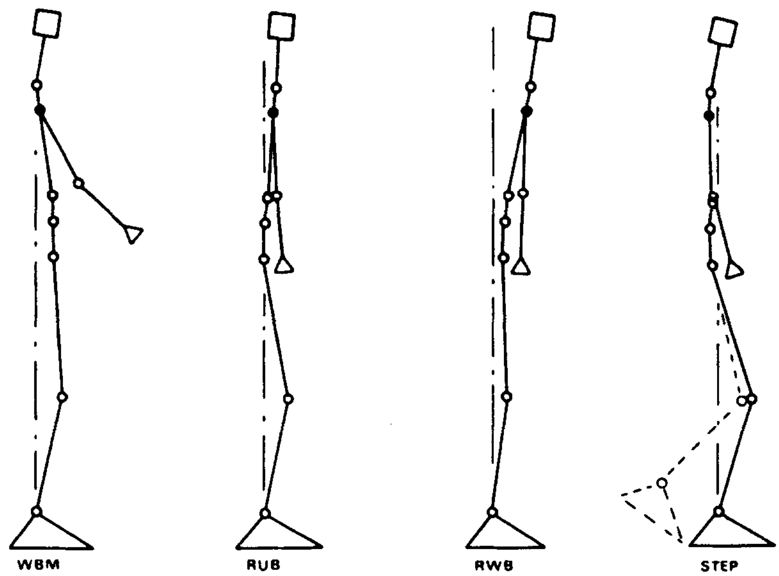


Fig. 6. Body configurations 500 ms after the onset of platform acceleration. Mean configurations relative to the feet (which were rotated about the heel up from the platform surface) are shown over the subjects exhibiting whole body movement (WBM), upper body rigid (UBR), whole body rigid (WBR) and stepping (STEP) responses.

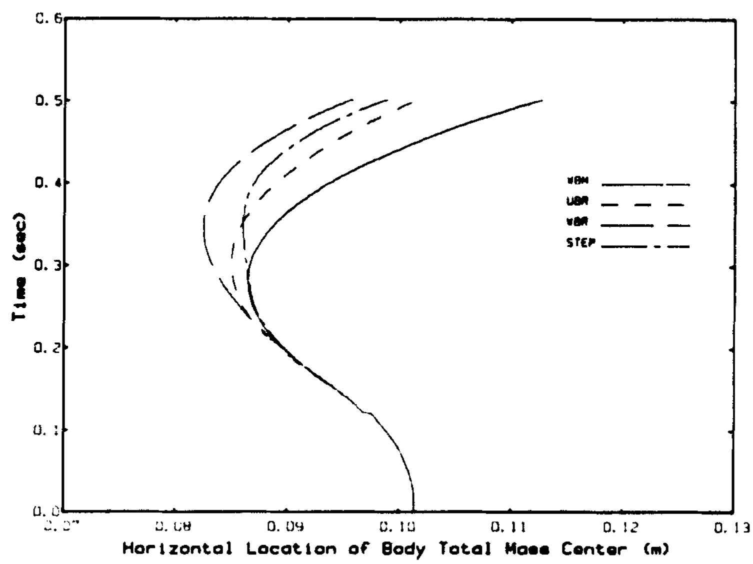


Fig. 7. Horizontal translation (m) of the body total mass center relative to the posterior-most point of the heels during the first 500 ms of response. Means are shown over the subjects exhibiting whole body movement (WBM), upper body rigid (UBR), whole body rigid (WBR) and stepping (STEP) responses.

pared to the others, required almost twice the ankle, knee and hip torques in the non-stepped leg because of the use of one leg support during the swing of the stepped leg.

The sensitivity analyses showed leg joint torques and total body mass center locations generally to be insensitive to the kinematic parameter changes. In response to 25% parameter changes, ankle torques changed at most by 6% and mass center locations by 5%. However, knee joint torques were more variable. For example, in WBR responses they changed by only 2%, while in WBM and UBR responses, they changed by up to 35%.

DISCUSSION

While the present results are preliminary in nature, no other comprehensive study of fall response for whole body biomechanics seems to have been reported. The study demonstrates that a combination of experimental measurements and model simulation analyses can be used to study comprehensively the biomechanics of reactions to impending falls. Thus, it may soon be possible to explain in terms of whole body biomechanics why the elderly fall so much more often than the young.

Some of the data reported, such as the myoelectric signal latencies and the joint ranges of motion, were observed directly, so there seems little question as to their validity. Angular acceleration data were obtained through least squares fitting of observed motion data and joint torque data were derived from the acceleration data through the biomechanical model analyses. Thus, the validity of those data can reasonably be questioned. However, the results of the sensitivity studies suggest that the predicted joint torques are probably correct at least as to order of magnitude.

In response to an anterior platform acceleration, simple mechanical analyses show that shoulder flexion tends initially to promote rather than to arrest the impending fall. However, the subjects routinely flexed their shoulders. Perhaps this was done to have the arms positioned to brake any subsequent impact, or to

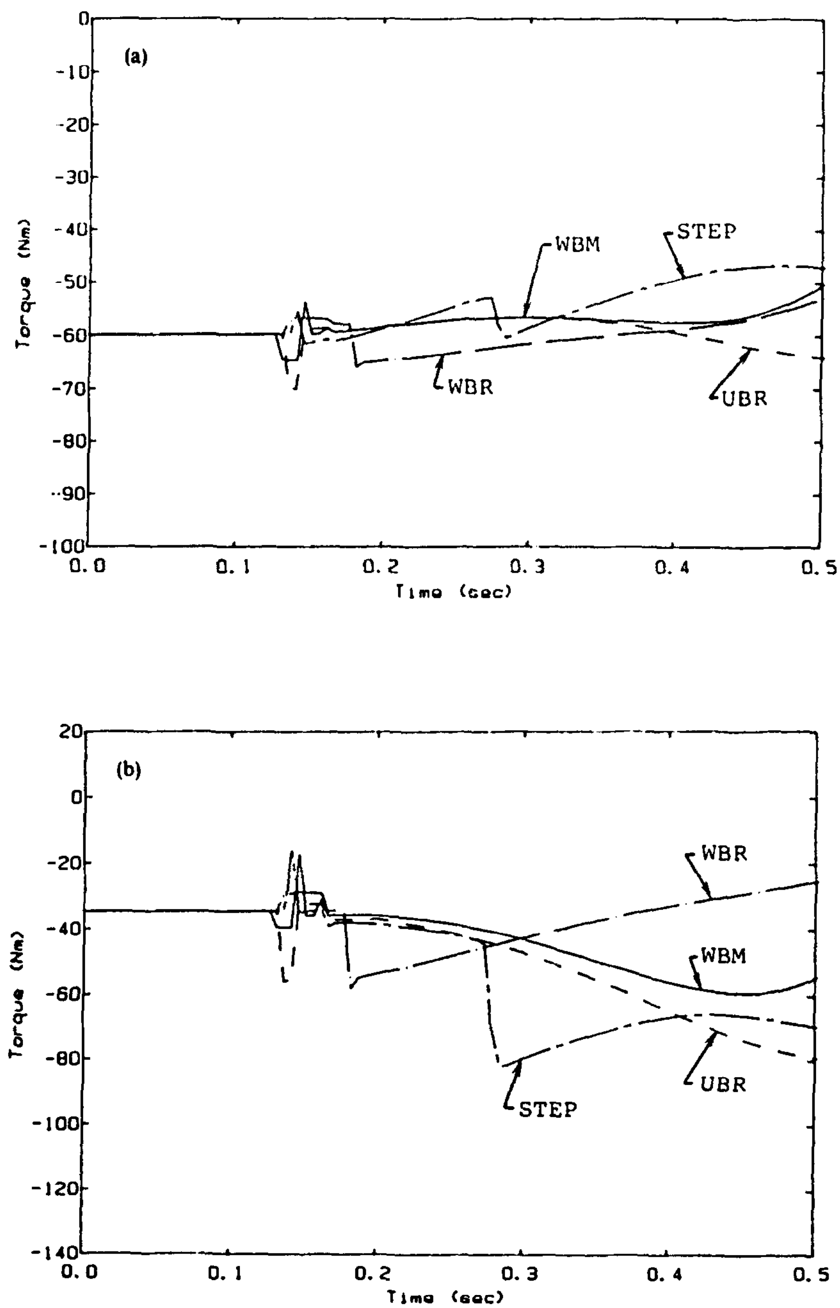
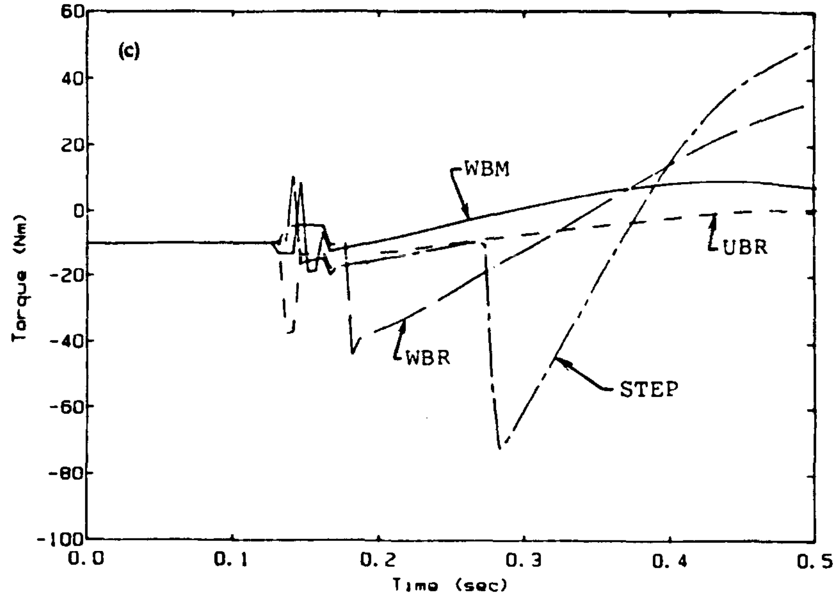


Fig. 8. (a, b)



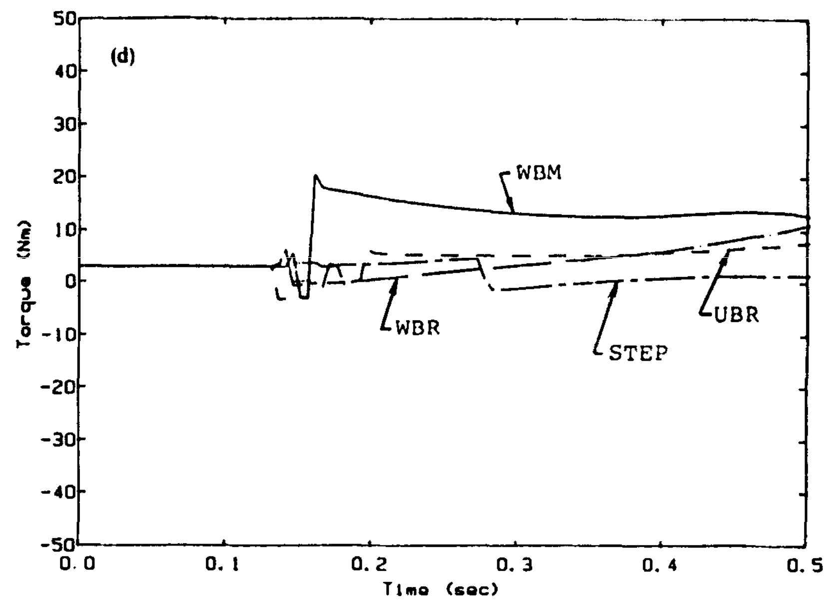


Fig. 8. (c, d) Calculated joint torques. Mean values are shown versus time for the ankles (Fig. 8a), knees (Fig. 8b), hips (Fig. 8c) and shoulders (Fig. 8d).

increase the total amount of inertia about the heel pivot, or to help keep the total body mass center within the area of foot support. The stepping responses clearly helped achieve the last goal, but compared to other types of responses, required greater leg muscle efforts to do so. In any event, the observed body segment relative motions seemed to be organized into only a few response patterns. In particular, mean muscle contraction latencies were always shortest at the ankles and longer at successively more superior body joints. The response efforts required from the leg muscles were the largest of any of the required muscular efforts.

The fall-promoting stimulus used here, floor acceleration, seldom would arise in natural situations. It probably best reflects conditions of a slip on an oily or icy surface. Nevertheless, it is a useful stimulus for laboratory studies of fall reactions because it can be fully controlled and is easy to model in biomechanical simulations of responses.

A number of improvements are called for in future studies of this type. A motion measurement system with high-enough acquisition rates would remove the need for waveform fitting of the motions, and hence remove the artefacts to which this led. Measurements of foot-floor interface reactions and quantification of myoelectric signals in terms of muscle contraction forces would serve as tests of biomechanical model validity. The latter would also quantify the extent to which co-contractions of antagonistic muscles occur, and so provide more accurate estimates of the extent to which the musculoskeletal system is stressed by fall reaction responses. It would be useful to monitor myoelectric responses in a larger number of muscles. Longer intervals prior to support surface deceleration would enable responses beyond 500 ms to be followed without the complications produced by those decelerations.

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