INFLUENCE OF HANDRAIL DESIGN ON POSTURAL STABILIZATION:

PILOT PHASE

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Summary

In previous studies of stairway handrails, data were derived from static experiments which characterized the influence of the handrail design on ability to generate stabilizing force. The work described here is the first phase of a series of studies that will examine the biomechanics of handrail use under dynamic conditions. These studies will be aimed at determining the demands placed on the handrail during stairway loss of balance, i.e. the magnitudes of force that must be generated to recover equilibrium. Furthermore, they will examine the scenario where the user is not initially contacting the rail, and must rapidly grab for it in order to recover equilibrium.

A novel experimental approach was developed to allow the simulation of stairway loss of balance in a safe and controllable manner. This approach involves the use of a moving-platform system to apply a postural perturbation to the experimental subject, who stands on the top tread of a three-step stairway mounted on the platform. The test handrail, which is instrumented to record the applied forces and moments, is also mounted on the platform. In each trial, the platform is gradually accelerated forward until it achieves a velocity that is representative of stairway gait, and is then controlled to decelerate suddenly, at an unpredictable moment in time. This deceleration causes the subject to pitch forward and downward, simulating an unexpected loss of balance during stairway descent. At the start of each trial, the subject stands with one leg extended forward, beyond the top tread, leaning against a backboard for stability. A cover placed over the stair tread forces the extended foot to miss the tread, as the subject falls forward, and to land on the tread below, thereby simulating an overstep.

The primary objectives of the pilot phase were to develop and test the instrumentation and experimental protocol, and to determine the influence and relative importance of specific task conditions: a) perturbation magnitude, b) stance leg (right or left), c) proximity to the handrail, and d) initial hand position (gripping the rail versus arms at sides). The study was also intended to address a more basic issue: is it even possible to generate a handrail grasping response with sufficient speed and accuracy to prevent a fall after losing balance on a stairway? Four healthy male subjects were tested, and each performed a total of 57 trials, under various task conditions. In the main series of trials, they were instructed to try to maintain balance by grabbing the handrail. A second task discouraged stepping by placing a small obstacle in front of the feet. In a third task, subjects were allowed to contact the handrail prior to the start of the trial.

The results demonstrated that an accurate grasping response, and sizeable stabilizing handrail force, can be generated very quickly in response to loss of balance. Furthermore, these stabilizing responses were clearly of functional significance, resulting in a marked reduction in the incidence of "falls", compared to trials where the handrail was absent. Quantitative results, in terms of the magnitudes and predominant directions of the forces, are summarized in the accompanying figure (page iii). The most consistent aspect of the force generation, seen across all task conditions, was the tendency to exert a forward "axial" force along the handrail. An unexpected finding was that this force often appears to be exerted through a pulling, rather than pushing, action, because of the posterior location of the grip relative to the body. Although most of the force components tended to increase with perturbation magnitude, the lateral forces appeared to be most dependent on whether or not a step was taken. Initial stance leg had few effects, but variation in proximity to the rail was found to influence the trajectory of the hand. In addition, there was a greater tendency to pull upward when the subject gripped the rail prior to perturbation. Both of these latter findings may have implications for handrail design. Overall, the present results support the feasibility of the new experimental approach. The results also indicate a number of specific ways in which the protocol can be streamlined, but suggest that trials to assess the influence of proximity to the handrail and initial contact with the rail should be included in any future studies. Modified protocols are proposed to assess potential limitations in simulating "real" stairway accidents, specifically the absence of downward body motion at perturbation onset and the fact that the experimental perturbations were not truly unexpected events.

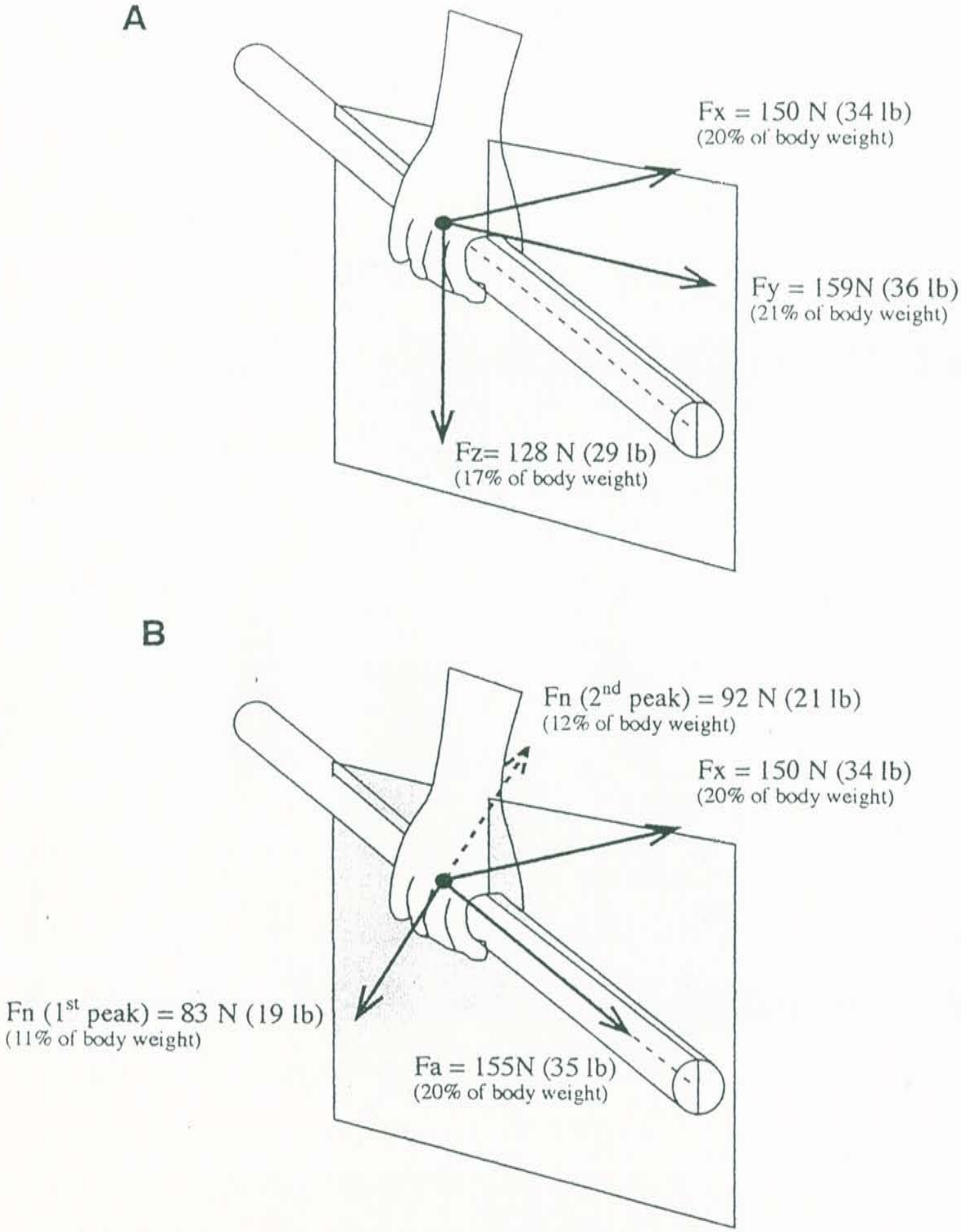

Figure S.1 - Forces acting on the handrail after grasping the rail to stabilize the body during experimentally-simulated stairway loss of balance; the predominant force components are shown, along with the average values of the peak force (recorded at the "medium" perturbation magnitude, corresponding to an average stairway gait velocity of 0.5m/s); panel A shows the Fy and Fz components that act in the sagittal plane (shaded); in panel B, the sagittal-plane force is represented by a component directed along the axis of the handrail (Fa) and a component perpendicular to the handrail (Fn).

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1. INTRODUCTION

1.1 Background

In previous biomechanical studies of stairway handrails, data were derived from static experiments which characterized the influence of the handrail design on the ability of the user to generate stabilizing reaction forces and moments[1-5]. The static force measurements were intended to provide comparative data relevant to the situation where the stairway user is gripping the handrail when loss of balance occurs.

The work described in this report is the first phase of a series of studies that will examine the biomechanics of handrail use under dynamic conditions. These studies will be aimed at determining the actual demands placed on the handrail during stairway loss of balance, i.e. the magnitudes of force that must actually be generated in order to recover equilibrium. Furthermore, they will examine the scenario where the stairway user is not initially gripping the handrail, and must rapidly grab for the handrail in order to recover equilibrium.

1.2 Purpose of the pilot phase

It is intended that future experiments will determine the biomechanical demands placed on stairway handrails across a wide range and large number of subjects. Future studies may also directly compare the efficacy of different handrail designs in generating stabilizing forces at the hand under dynamic conditions. The pilot phase described in this report was intended to develop the instrumentation and testing protocols that will be needed for these future experiments and to provide preliminary data that will be needed to design these experiments. These specific objectives are outlined in the section that follows.

Although the pilot phase was limited to a very small number of subjects, it was also anticipated that the pilot results could, in themselves, be of potential importance to researchers and practitioners who are interested in stairway safety, particularly since so little research has been done in this area. First and foremost, the collected data allowed us to address a very basic question: is it actually possible to prevent a stairway fall by grasping a handrail, at perturbation magnitudes that are sufficient to evoke loss of balance? In addition, we were able to examine whether the extent of initial contact with the handrail affects the ability to prevent loss of balance (i.e. gripping the handrail versus no initial contact). Both of these issues have important implications for the way that handrail design is used to promote stairway safety. Finally, in the absence of any previous data on handrail demands, it was anticipated that the pilot results might be useful to handrail designers and building code officials in providing some indication of the range of handrail forces that can be expected to occur during stairway loss of balance.

1.3 Specific objectives of the pilot phase

1. Equipment, instrumentation and software development:
   1) Design and construct the equipment needed to simulate stairway loss of balance.
   2) Instrument the handrail to allow the applied forces and moments to be measured accurately for all three directional axes.
   3) Develop software to control the perturbation system and to perform the data acquisition

during experiments.

2. Development of the testing protocol:
   1) Determine appropriate perturbation parameters for inducing loss of balance.
   2) Determine the extent to which the protocol is effective in preventing "learning" effects.
   3) Determine the extent to which the protocol is successful in preventing anticipatory activation of arm muscles.

3. Collect information needed to design future experiments:
   1) Determine whether it is possible to maintain balance using the handrail alone (without stepping) at perturbation magnitudes that would otherwise be sufficient to cause loss of balance. If it is possible to balance without stepping, determine the extent to which stepping (versus not stepping) influences the demands placed on the handrail.
   2) Determine which task conditions are most important, by analyzing the extent to which the demands placed on the handrail are influenced by:
      a) perturbation magnitude
      b) stance leg (right or left)
      c) lateral distance from the handrail
      d) initial contact (gripping the handrail versus arms at sides)
   3) Determine the variability in the handrail force measurements (these data will be needed for future sample size calculations).

2. METHODOLOGY

2.1 Overview of the experimental approach

Experimental studies of stairway falls face a fundamental methodological problem: how to simulate the biomechanical events of the fall, and the unpredictable nature of these events, in a safe and ethical manner? Most proposed approaches have involved the use of stairways that incorporate "trick" steps. Proposed safety measures have ranged from the use of a safety harness, wearing of protective padding and headgear, and even the use of professional "stuntmen" as experimental subjects[6-8]. The fact that few, in any, such studies have ever been completed attests to the problems inherent to these approaches. Although a harness may be successful in limiting the extent of the fall (and thereby preventing serious fracture injuries), there remains a significant risk of sustaining soft tissue injury (e.g. contusions, abrasions, sprains, joint injuries). Padding may help to prevent some of these injuries, but may tend to interfere with the motion that is of interest. The generalizability of results obtained from "stuntmen" subjects is questionable. A second problem in using "trick" stairs relates to issues of predictability, i.e. will subjects begin to alter their gait pattern in anticipation of experiencing the perturbation? A third problem relates to issues of fatigue: how many trials can be performed before the subject begins to tire appreciably?

In an attempt to resolve these problems, we have developed a new approach to simulating the biomechanical events of the stairway fall. Specifically, we have chosen to simulate an "overstepping" of the tread during stairway descent (i.e. the leading foot either misses the tread completely, or slips off the tread, in either case landing on the next tread below). This scenario was selected because falls occurring during descent tend to result in the most serious injuries, and it would appear that oversteps are one of the more common causes of these falls[9].

Our approach involves the use of a moving platform system to apply a postural perturbation to the experimental subject. The subject stands on the top step of a small mock stairway which is mounted on the moving platform. The platform is accelerated in a forward direction until it achieves a velocity that is representative of stairway gait. During this interval, the subject is supported by a backboard, which serves to prevent loss of balance. After achieving the desired velocity, the platform is then controlled to decelerate suddenly, at an unpredictable moment in time. This deceleration causes the subject to pitch forward and downward relative to the platform, thereby simulating a loss of balance during stairway descent. In order to further simulate the biomechanics of a stairway fall, the subject stands, initially, with one leg extended forward, beyond the top step, leaning against the backboard for stability. To simulate an overstep, a cover is placed over the next stair tread, thereby forcing the extended foot to miss the tread and to land on the step below (see Figures 2.1 and 2.2).

Since the mock stairway comprises only three steps, there is a very limited distance that the subject can fall. Moreover, a thick foam "crash pad" is placed at the foot of the stairway. Since there is no actual stairway gait prior to the perturbation, we have eliminated the concern that variations in gait pattern might confound the results. Unpredictability is easily achieved by varying the time of onset of the platform deceleration. Since there are no traverses of an actual stairway, fatigue is less of a problem and larger numbers of trials and test conditions can be tested.

To avoid evoking arm responses prior to deceleration onset, the initial platform acceleration is small in magnitude and has a smooth profile (an acceleration "ramp", with a small constant jerk).

In addition, the backboard mounted behind the subject prevents the initial acceleration from perturbing the posture of the body. To further minimize anticipatory responses, the duration of the acceleration ramp is varied unpredictably. In addition, a wide range of perturbation magnitudes (including small perturbations) are tested so as to discourage the preplanning of "default" responses[10]. To prevent subjects from focusing their attention on the balancing task, they are instructed to perform a secondary mental arithmetic task during the testing.

2.2 Apparatus and instrumentation

The mock stairway comprised three treads having a 184mm (7.25in) rise and a 241mm (9.5in) run; a 32mm (1.25in) nosing was added to the tread. The bottom tread was approximately 50mm (2in) above the surface of the platform. The step dimensions were selected to represent a typical residential stairway. As indicated earlier, the middle step was covered with a curved piece of polished vinyl in order to prevent the foot from landing on this step (see Figures 2.1 and 2.2). The vinyl cover was polished and shaped so that the foot would tend to slip down to the next stair tread, if in fact, the foot did initially land on the cover. Mounted across the back of the stairway was a backboard, which projected 2m (80in) above the top stair tread. The mock stairway and backboard were 70cm (28in) wide. A foam rubber "crash pad" (86cm x 90cm, 37cm in height; 34in x 35in x 15in) was mounted in front of the bottom stair tread, and a foam rubber barrier (45cm in height; 7cm thick) was placed on top of the crash pad, across the edge closest to the stairway, to discourage subjects from stepping onto the crash pad (see Figures 2.1 and 2.2).

In order to simulate the stairway loss of balance, the mock stairway was mounted on a computer-controlled, motor-driven moving-platform system that had been developed previously for studies of postural control[11]. Characteristics of particular relevance to the current study include the large size of the platform surface (2m x 2m), the relatively large range of motion (0.6m in any horizontal direction) and the robust accelerational capabilities (up to $10m/s^2$; frequency response flat to 5Hz; maximum velocity of 2m/s). An accelerometer recorded the platform acceleration to within $\pm$ $0.0025m/s^2$, and a linear potentiometer recorded platform displacement to within $\pm$ 0.0001m.

Mounted on the platform were three biomechanics force plates: two strain-gauge models (Model OR6-7-2000 and OR5-6-1, AMTI; Newton, MA) and one piezoelectric model (Model 9281, Kistler; Amherst, NY). All force plates recorded all six force and moment components applied to them. The handrail mounting system, comprising a base and two posts (1.4m, or 55in, apart) was bolted to the two AMTI force plates. This mounting system was designed to accommodate a wide range of different handrail cross-sectional shapes and sizes, and to allow the height of the handrail to be adjusted in 5cm (2in) increments over a range of 76cm (30in) to 107cm (42in). The mock stairway was mounted on the Kistler force plate, to allow the reaction forces on the subject's feet to be measured (these data were collected but are not analyzed in the present report). In addition, the bottom stair tread was covered by a thin plate, mounted on tapeswitches, in order to record the timing of foot contact.

Bipolar Ag-AgCl surface electrodes (8mm diameter, 25mm between centers) were used to record electromyographic (EMG) activity in five muscles of the right arm (deltoid, triceps, biceps, flexor digitorum and extensor digitorum) and in four muscles of the left leg (tibialis anterior, medial gastrocnemius, rectus femoris and medial hamstrings). In connecting the electrodes to the preamplifiers, great care was taken to secure the wires in such a way that the motion of the subject was not, in any way, impeded. The EMG preamplifiers and filters were suspended over the moving platform, to the side of the subject, by a cable from the ceiling of the laboratory (mounting this instrumentation on the platform itself was found to create motion artifacts, due to the large jerk associated with platform deceleration). A fiber-optic cable transmitted the signals to the main signal processing unit and data acquisition system. Although a variety of muscle sites were recorded for the purposes of future exploratory analyses, the primary purpose of recording EMG, with regard to the present report, was to allow detection of anticipatory grasping responses beginning prior to onset of platform deceleration. Our previous studies[12,13] have shown, and the present data confirmed, that the earliest arm activation associated with grasping movements occurs in the shoulder abductor (deltoid); therefore, this muscle was used as the primary indicator of anticipatory activity. In trials where the hand was gripping the handrail prior to perturbation, the deltoid was frequently inactive; in these trials, the finger/wrist extensors (extensor digitorum) were found to give the most consistent indicator of the earliest arm activation.

Two synchronized PC-based data acquisition systems were used to sample the force plate, accelerometer, potentiometer, tapeswitch and EMG data. The EMG signals were sampled at a rate of 1000 Hz; all other signals were sampled at a rate of 200 Hz. Signals were low-pass filtered (second order Butterworth filters: -3dB at 500 Hz for EMG, -3dB at 10 Hz for other signals) prior to sampling in order to prevent aliasing. In addition, the EMG signals were high-pass filtered (second order, -3dB at 10 Hz) in order to remove motion artifacts.

A commercial video-based motion-analysis system was used to record the motion of reflective markers placed on the body of the subject (Peak Performance Technologies, Inc.; Englewood, CO). This system uses SVHS recorders to store the video signals, which are then analyzed offline by a semi-automated computer-controlled digitizing system (the operator intervenes only when markers are lost from camera view). Five high-resolution CCD cameras (shuttered at 1/500s) were used in the current study: two cameras viewed the front of the staircase (each at an angle of about 45 degrees to the axis of platform motion), one camera provided a side-view of the stairway (viewed from the handrail side) and two cameras were placed overhead. Each marker must be viewed by at least two cameras in order to determine the three-dimensional coordinates. In practice, some markers were obstructed from some camera views, during the course of the movement; hence, the need for the extra cameras. The cameras were synchronized ("genlocked") with respect to each other and SMTE time code was recorded on the audio tracks of the videotapes. In addition, to allow synchronization with the sampled signals (force plate, EMG, etc), timing pulses were recorded on the videotape at onset of data sampling, platform acceleration and deceleration. The effective sampling rate of the video system was 60Hz.

A total of 23 reflective markers (styrofoam balls, 20-40mm in diameter, covered with reflective tape) were placed on the subject, using double-sided adhesive tape. The following marker locations were used: dorsum and ankle (lateral malleolus) of each foot, pelvis (left and right anterior superior iliac spines), sternum, left and right shoulders (acromion), head (chin, left and right ears, forehead), and left and right wrists. Additional markers were placed on the shank and thigh of the left leg (it was not feasible to record the right leg, because the handrail system tended to obscure the markers), and on the upper arm, forearm and hand (thumbnail and nail of middle finger) of the right arm. Reference markers were placed on the moving platform and on the handrail. In placing the markers on the subject, care was taken to ensure that the markers (particularly those placed on the right arm and hand) did not interfere with the motion of the subject or the ability to grasp the handrail. The large array of marker sites was used to allow the possibility of performing detailed kinematic analysis. Such analyses are beyond the scope of the present report. Instead, we have focused primarily on the right wrist marker, which was used to define the handrail grip location, and the sternum marker, which was used as an indicator of the motion of the whole body.

2.3 Testing of instrumentation calibration

Calibration coefficients for the force plates were provided by the manufacturers. The accuracy of the calibration was verified by performing a series of static loading tests. A loading frame was constructed to perform these tests. This was simply a steel support frame that bridged the force plates. Mounted on the frame was a ball bushing, which supported a vertical post and platter assembly. To apply vertical loads to the force plate, the frame was moved so that the vertical post was aligned with the desired location on the force plate and calibrated lead weights were then stacked on the platter. A grid of calibration locations was marked on each force plate surface. The spacing between the grid locations was 15-20 cm (6-8in), resulting in a total of 21-24 loading locations for each of the AMTI force plates. At each location, eight loads were applied ranging, in approximately 110N (25lb) increments, from 70 to 900N (15-200lb). These vertical loads resulted in both positive and negative moments about the x-axis (medio-lateral, m-l) and y-axis (antero-posterior, a-p), ranging from about 15 to 550Nm (10-400ft-lb). To apply horizontal loads (and moments about the vertical, z, axis), a calibrated load cell and turnbuckle assembly was connected, by wires, between attachment points on the moving platform surface and the force plate surface (two load locations were used: one near the center of the force plate and the other offset by approximately 30cm, or 12in). By adjusting the turnbuckle, tensile loads were applied to the force plates. Five horizontal loads were applied at each of the four locations, along each of the positive and negative a-p and m-l axes. These loads ranged, in approximately 110N (25lb) increments, from 25 to 475N (5-105lb). The corresponding moments about the vertical axis ranged, in absolute value, from 15 to 150Nm (10-100ft-lb). The results of these calibration tests are summarized in Tables 2.1a and 2.1b.

In addition to the above, the calibration of the complete handrail system, mounted on the two AMTI force plates, was verified by applying loads to the handrail itself, using a spring-scale and a torque wrench. The loads applied in these tests could not be controlled as accurately as in the separate force-plate calibrations described above, due to inaccuracies in determining the exact magnitude, location and direction of the applied load. Nonetheless, these tests were useful in verifying that the algorithms developed to calculate the overall forces and moments applied to the handrail (using the forces and moments measured by the individual force plates) were yielding results that were of appropriate magnitude and sign (polarity).

A calibration frame was used to calibrate the video motion-analysis system. This frame comprised 17 reflective balls (25mm in diameter) which were mounted on rods that projected outward from a central mounting block, which was mounted on a tripod. This frame, which was supplied by the manufacturer of the system (Peak Performance, Inc.), had been surveyed so that the three-dimensional coordinates that define the relative position of each ball are known to a very high degree of accuracy (0.1mm). The calibration frame was placed on the moving platform, so that the calibration volume encompassed both the mock stairway and handrail, and the axes of the frame were aligned with the axes of motion of the moving platform by dropping plumb bobs from the frame to the platform surface. The frame was videotaped for approximately 5min, and the positions of the reflective markers on the frame were digitized. These data were then used to calculate the parameters of the Direct Linear Transform, which is the method used by the motion-analysis software to transform the digitized data into three-dimensional coordinates[14]. The total volume occupied by the calibration frame was approximately 75cm (30in) in height, 80cm (32in) in the medial-lateral (m-l) direction, and 75cm (30in) in the anterior-posterior (a-p) direction. Over this viewing volume, the root-mean-square errors in the a-p, m-l and vertical coordinates of the calibration markers were 3.7mm (0.15in), 4.8mm (0.2in) and 5.0mm (0.2in), respectively.

2.4 Perturbations

The platform motion was intended to accomplish two objectives: 1) the initial acceleration would cause the body to achieve a velocity and momentum that is representative of stairway gait, and 2) the deceleration would cause the body to pitch forward and downward relative to the stairway so as to simulate a loss of balance during stair descent. As the platform motion is arrested, the center of mass of the body begins to move forward relative to the stairway. Furthermore, as the center of mass moves past the supporting foot, the force of gravity begins to pull the body downward.

The deceleration was performed as rapidly as possible, within the capabilities of the platform system, so as to maximize the destabilizing effect. The duration of the deceleration pulse was approximately 250ms. Attempts to shorten this duration tended to cause a significant degree of overshoot, i.e. the platform actually moved beyond the desired stopping point and then returned back to the desired location.

One important consideration, in designing the perturbations, was to minimize any destabilizing effects during the acceleration phase, so as to enable the subject to maintain the desired starting position (standing on one leg, with the other leg extended forward). The backboard played an important role in meeting this objective, by preventing the backward sway that the forward acceleration of the platform would otherwise tend to induce. To further prevent destabilization, or evoking of postural reactions, the acceleration phase of the perturbation was designed to be executed as gradually and smoothly as possible, i.e. with minimal jerk. Another important consideration was to make the timing and magnitude of the perturbation, due to the platform deceleration, unpredictable to the subject.

Initially, we experimented with waveforms having an interval of constant velocity prior to onset of deceleration. The desired velocity was achieved by means of a smooth polynomial acceleration waveform. However, it was found that the decrease in acceleration that occurred as the platform reached the desired velocity plateau was clearly perceptible to the subject, and, moreover, tended to elicit a stabilizing reaction. To prevent this, we instead elected to use an acceleration ramp, i.e. a constant rate of change in acceleration (jerk) to build up to the desired velocity; the deceleration pulse was administered immediately once the target velocity was reached. The slope of the acceleration ramp was identical for all perturbations (jerk = $0.5 m/s^3$) and the only change in jerk (other than at the start of the platform motion) occurred at onset of deceleration. By varying the duration of the acceleration phase, the peak velocity could be varied; however, since the subjects would not be able to predict this duration, they would not be able to predict either the peak velocity or the time of deceleration onset. In other words, the subject experienced a platform motion that gradually increased in speed (and this gradual acceleration always occurred at the same rate), but had no idea at what point in time the platform motion would be stopped.

Three perturbation magnitudes ("small", "medium" and "large") were used in the experiment. The corresponding maximum velocity values were selected to represent a wide range of stairway gait: 0.25, 0.5 and 0.75m/s (average stairway gait velocity is about 0.5m/s; the "small" and "large" values are more than two standard deviations from the average[15]). Actual recordings of platform acceleration, velocity and displacement are displayed, for each of the three perturbation magnitudes, in Figure 2.3. The actual perturbation characteristics, as determined from the platform accelerometer and potentiometer signals recorded during each experimental trial, are listed in Table 2.2.

2.5 Test conditions and procedures

The same handrail was used in all tests: 864mm (34in) in height (measured from the leading edge of the tread, in the plane of the riser, to the top surface of the handrail), circular cross-section, 51mm (2in) in diameter. These parameters are close to the upper limits, for height and size respectively, of most building codes, at the present time. The handrail was constructed of aluminum tubing (wall thickness of 3.5mm or 0.125in), and was painted "flat black" (Tremclad Rust Paint) to prevent reflections that would interfere with the digitization of the reflective body markers. The matte finish also served to minimize slippage of the hand.

The mock stairway comprised three treads having a 184mm (7.25in) rise and a 241mm (9.5in) run; a 32mm (1.25in) nosing was added to each tread. The bottom tread was approximately 50mm (2 in) above the surface of the platform The step dimensions were selected to represent a typical residential stairway. As indicated earlier, the middle step was covered with a curved piece of polished vinyl in order to prevent the foot from landing on this step (see Figures 2.1 and 2.2). The vinyl cover was waxed and polished so that the foot would tend to slip down to the next stair tread, if in fact, the foot did initially land on the cover.

The handrail was mounted to the right of the subject, on the moving platform, at the same pitch as the mock stairway (37 degrees). Two lateral displacements, between the subject and the handrail, were tested: 1) a *"close"* position (32cm between the midline of the body and the center-line of the handrail), and 2) a *"far"* position (61cm from body midline to handrail center-line). The close position represents the average preferred lateral position recorded in our previous studies[1-5]. The far position is an estimate of the maximum lateral displacement at which both a "5th percentile female" and a "95th percentile male" could reach the handrail without leaning sideways[16-18]. It also corresponds, approximately, to maximum lateral displacements observed during field studies of traffic patterns on stairways having a 1.5m spacing between parallel handrails, the maximum spacing permitted by some building codes[16-18]. Markings were placed on the stair tread to indicate the two stance positions.

Prior to the start of each trial, the subject was instructed to stand on the top step of the mock stairway, with arms relaxed at his sides. Initially, the subject stood with heels together, centered on the appropriate stance position marking, and the back of the heels touching the backboard. Leaning against the backboard for stability, the subject was then asked to shift his weight onto either the left or right foot and to extend the other foot forward past the edge of the stair tread, with the back of the heel resting lightly against the nosing (see Figure 2.1). The subject was instructed to keep his arms relaxed at his sides, initially. To prevent variation in visual cues, the subject was instructed to look straight ahead at a target mounted, at the subject's eye level, on the wall of the laboratory, approximately 5m (16ft) from the starting position of the platform.

The subject was told that, in each trial, the platform would begin to move slowly and would gradually accelerate for a few seconds before stopping suddenly. It was emphasized that it was the stopping of the platform that would cause the subject to lose balance, that the point at which the platform stopped would be varied at random, and that the subject should endeavour to hold his initial position until he actually started to lose balance. To distract the subject from thinking about his response to the perturbation, he was asked you to count backward by 7's, out loud, as fast as possible, and to keep counting until the platform stopped moving. The starting number for the counting task was varied between trials, to minimize practice effects and thereby preserve the level of cognitive "load".

2.6 Protocol

Each subject participated in a single testing session. After three initial familiarization trials, three blocks of trials were performed (*"main experiment"*). Each of these blocks comprised 12 trials, performed in random order: one trial at each of three magnitudes of perturbation and four starting conditions (right or left stance leg, lateral displacement of 32cm or 61cm). During these trials, the subject was instructed to try to maintain balance by grabbing the handrail, and to avoid stepping if possible. At the start of each trial, the subject stood on the specified stance leg, with the other leg extended in front, as described previously. Arms hung relaxed at the sides.

The main experiment was followed by a block of trials in which an obstruction was used to further discourage stepping (*"feet-obstructed trials"*). In these trials, the subject stood on both feet (heels together, feet at a comfortable angle), and a piece of foam rubber was placed immediately in front of the toes. This piece of foam was placed on top of the tread cover that was mounted over the intermediate step; the top of the foam rubber was approximately 5cm (2in) higher than the surface of the top step. This arrangement was designed with safety in mind, i.e. subjects were well able to step through, or over, the obstacle, if necessary. Six trials were performed in random order: one trial at each of the three magnitudes of perturbation and two starting conditions (lateral displacement of 32cm or 61cm).

Following the feet-obstruction trials, a block of trials was performed in which the subject was instructed to grip the handrail, prior to the start of each trial (*"hand-on-rail trials"*). The preferred grip location was first determined and marked on the handrail, so that the same location was used in all of the trials. The subjects were instructed not to exert any force on the handrail until they started to lose their balance. All trials were performed at the "close" stance position (lateral displacement of 32cm); it was not possible to reach the handrail from the "far" position without leaning sideways. Six trials were performed in random order: one trial at each of the three magnitudes of perturbation and two starting conditions (right or left stance leg).

The session concluded with one block of trials performed in the absence of a handrail (in the very first subject, these trials were performed during a second testing session performed approximately 2 weeks later). These trials were included to provide some baseline data for comparison with the handrail responses, i.e. to allow the stabilizing contribution of the handrail to be assessed. Six trials were performed in random order: one trial at each of the three perturbation magnitudes and two stance-leg conditions (right or left).

The total number of trials per session was 57. Each trial took approximately one minute to complete, and 5-minute seated rest periods were allowed between trial blocks. About 30 minutes was required for subject preparation, i.e. placing of markers and electrodes, at the start of the session, and another 30 minutes was needed at the end for removal of markers and electrodes and collection of anthropometric data. The total duration of each testing session ranged from 2.5 to 3.5 hours. See Appendix A for full details of the protocol.

2.7 Subjects

Four healthy young-adult males were tested. All subjects were right-handed. In response to a questionnaire administered by the Research Assistant, none of the subjects reported any significant neurological, sensorimotor or musculoskeletal disorders/deficits, or use of drugs or medications that might affect postural balance, the ability to descend a stairway or the ability to grasp a handrail. The characteristics of the subjects are summarized in Table 2.3.

2.8 Analysis

The forces exerted by the hand on the handrail were calculated by summing the forces measured by the two AMTI force plates. In order to calculate the pure moments exerted by the hand, it was also necessary to know the location of the loading point relative to the force plates, in order to subtract the moment exerted on the force plates by the handrail force from the total moment measured by the force plates. The grip location was estimated from the digitized coordinates of the reflective marker placed on the back of the right wrist, at the base of the hand. The a-p coordinate of the grip force was assumed to be coincident with the a-p location of this marker. In determining the m-l and vertical coordinates of the loading point, it was assumed that the force passed through the center-line of the handrail.

It should be noted that the accuracy of the pure-moment calculations is limited by the accuracy with which the moment due to the handrail force can be calculated. The latter is influenced by errors in the determination of the grip location. Moreover, small errors in the determination of the handrail force itself can have a relatively large effect on the moment calculation, because of the large moment arms that are involved. For example, a 5N (1lb) error in a-p force could lead to a 5Nm error in the determination of the pure moment about the m-l axis, because the force acts at a moment arm of about 1m relative to the force plates.

The force and moment calculations were corrected for inertial artifacts, due to the acceleration of the force plates, by subtracting the forces and moments measured when the apparatus alone (plus the "deadweight" of the subject, holding on to the backboard) was subjected to the perturbation. In practice, these corrections had little effect on the determination of the peak forces and moments, because these peaks almost invariably occurred after the platform had stopped moving. Example data, illustrating the effects of the inertial correction, are provided in Figure 2.4.

Prior to determining the peak forces and moments, the force plate signals were low-pass filtered, digitally (fifth order Butterworth filter, -2dB at 4Hz, -20dB at 6Hz; the data were time-reversed and passed through the filter twice, in order to achieve zero phase lag), in order to remove high-frequency artifacts arising from excitation of the measurement system. This excitation arose from inertial forces associated with high-frequency components present in the platform deceleration pulse, and from the impact loading created by the grabbing response, which sometimes caused the system to "ring" slightly at its natural frequency. Impact loading tests of the handrail revealed that the natural frequency of the combined handrail and force plate system ranged from 10 to 20Hz, depending on the loading direction. The filter cut-off of 4Hz was selected to eliminate the higher frequency noise associated with the natural frequency of the measurement system, while preserving the frequency content of the biomechanical forces. Muscle-generated forces, as well as passive forces associated with motion of the body, generally have a frequency content that is less than 3Hz[19].

For each trial, the peak forces and moments acting <u>on</u> the handrail were determined for each of the three directional axes, and in both directions along each axis (see Figure 2.5). The resultant was also determined. The sagittal-plane forces and moments were each represented using: 1) inertial axes (i.e. vertical and a-p components), and 2) handrail axes (i.e. an "axial" component tangential to the longitudinal axis of the handrail and a "normal" component perpendicular to the longitudinal axis). The former representation allows the stabilizing effect of the force to be related to the a-p, m-l and vertical motion of the center of mass, whereas the axial/normal representation relates more closely to the two main biomechanical aspects of the grip (i.e. frictional shear versus contact pressure).

In searching the force/moment records for the peak values, the search began at time of initial contact with the handrail and terminated when the falling motion of the body had been arrested (as estimated from the video recordings). In cases where the subject grasped the handrail with the left hand, the search was terminated at that point, to ensure that the measured forces pertained to single-handed grasping (right hand) only.

The primary variables of interest, in the current study, were the peak forces, as well as the timing of these peaks. Peak values of the pure moments, generated at the hand, were also analyzed. To facilitate comparison between individuals of differing size, the force variables were normalized by dividing by body weight. In addition to the force and moment variables, we also analyzed the time of initial arm-muscle activation, the time required to contact and gasp the handrail, and the location of the hand at the time when a full grasp was achieved (hand location was indicated by the reflective marker on the wrist). The net effect of the perturbation, and the stabilizing reactions, on the motion of the body was characterized in terms of the peak velocity of the reflective marker placed on the sternum. All timing variables were defined relative to the onset of the platform deceleration, i.e. change in jerk exceeding $10 m/s^3$.

For each variable of interest, a four-way repeated-measures analysis of variance (ANOVA)[20,21] was performed to determine the influence of: 1) perturbation magnitude, 2) lateral stance position (close or far), 3) stance leg (right or left) and 4) repeated trial number (round 1, 2 or 3). Note that the repeated-measures ANOVA is actually a mixed-effects model: the four factors listed above constitute *fixed* effects which are "repeated" within subjects, whereas subjects are treated as *random* effects (in the sense that they are intended to represent a random sample from a larger population). This is also known as a "randomized complete block design", with subjects considered to be the blocks.

The assumptions underlying the ANOVA (i.e. normality, uniformity of variance and independence of residuals) were checked through examination of the model residuals[21]. In cases were violations were suspected, the data were log- or rank-transformed[22], and the analyses repeated. In addition to the above, a second set of analyses was performed to compare trials where the subjects were able to maintain balance without stepping to those where stepping occurred. Due to the limited size of the data set, it was not possible to include all of the other factors in these analyses (perturbation magnitude and stepping/non-stepping were the only factors included).

The above analyses were limited to the data collected during the main experiment. Additional ANOVAs were performed to compare the main-experiment trials to the feet-obstructed and hand-on-rail trials. Due to the small numbers of trials in the latter task conditions, the only factors that were included in these analyses were task condition and perturbation magnitude.

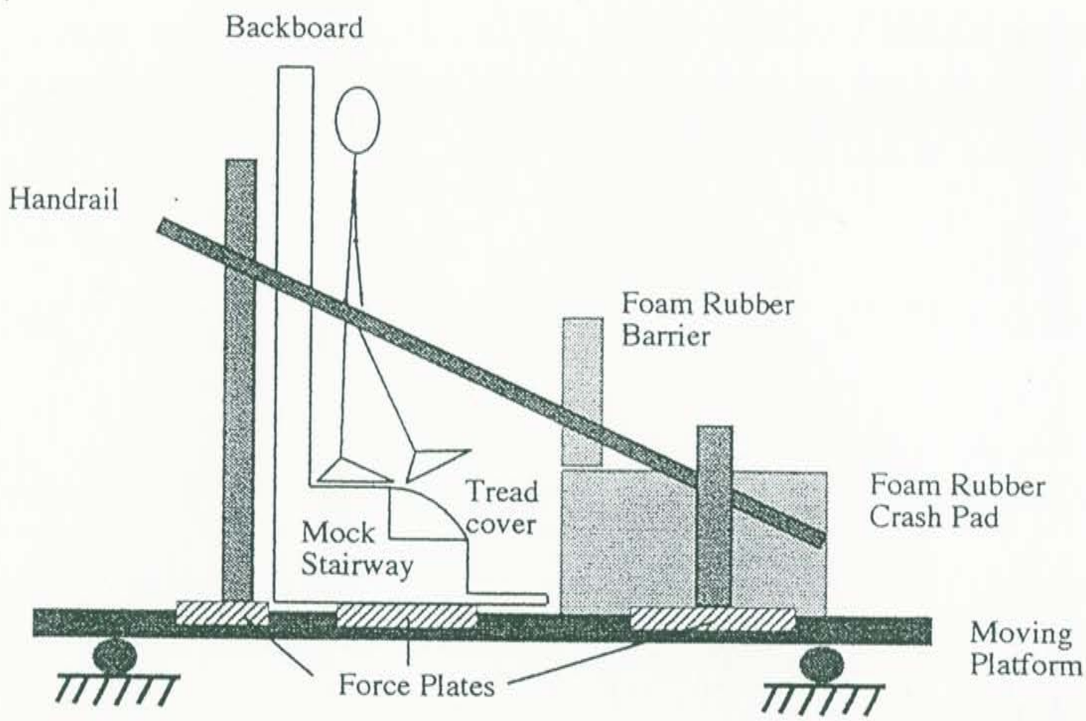


Figure 2.1 - Schematic drawing of the experimental apparatus.

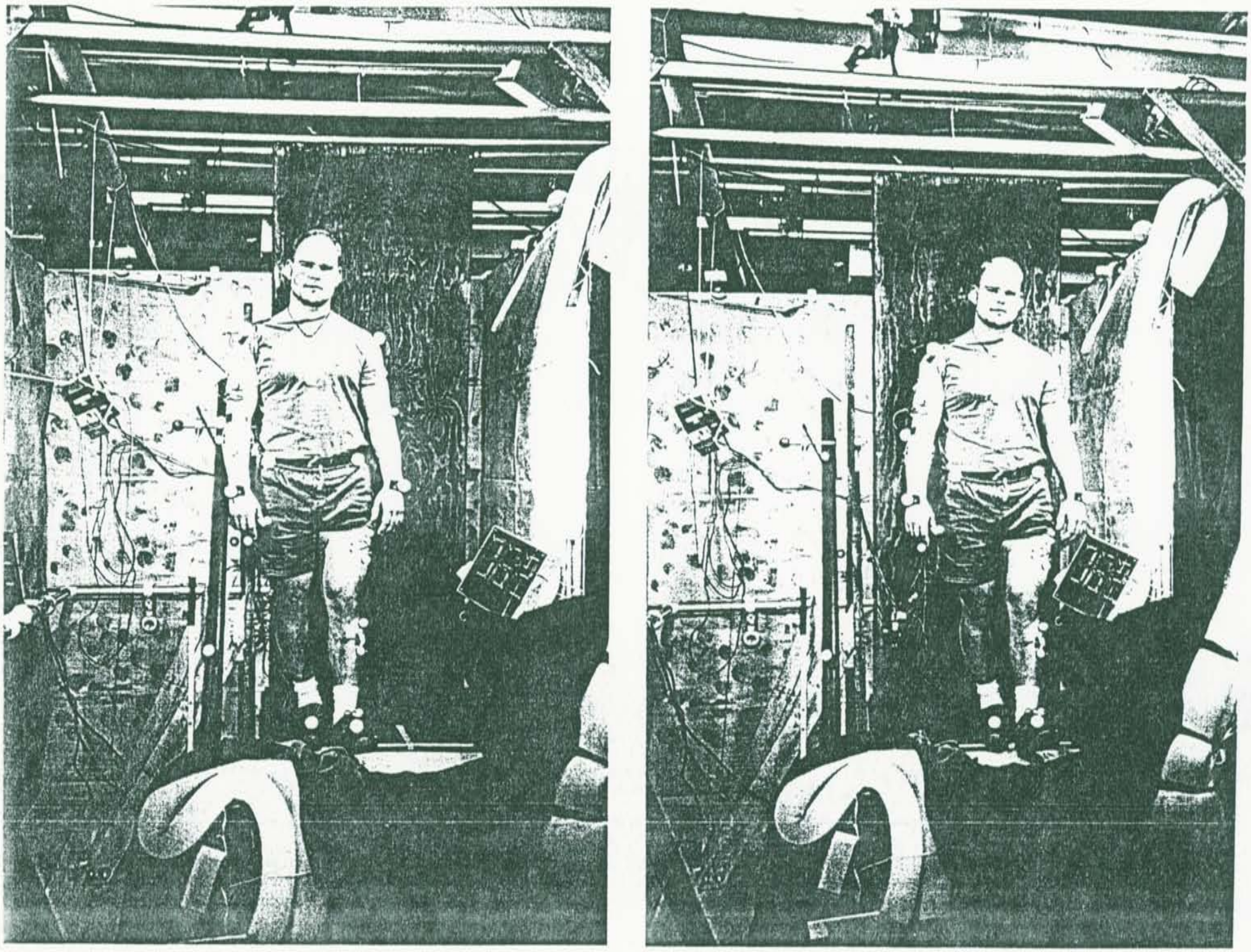
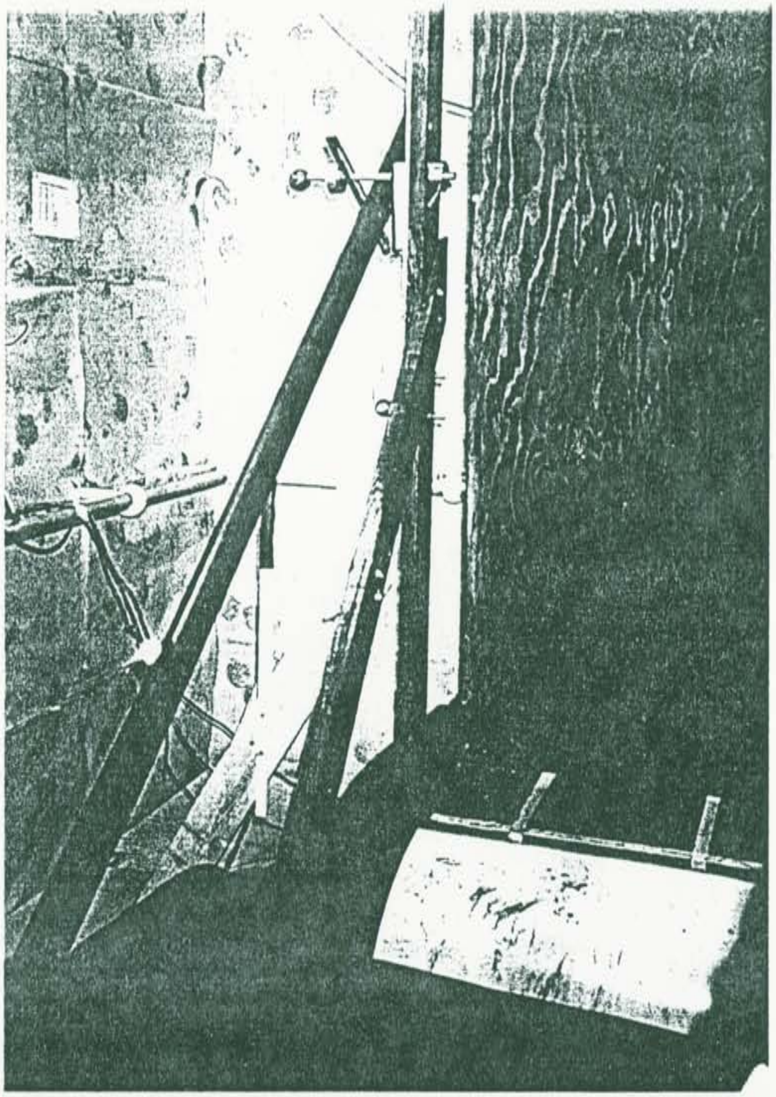


Figure 2.2a - Photographs of the experimental apparatus, illustrating the "near" and "far" stance positions.



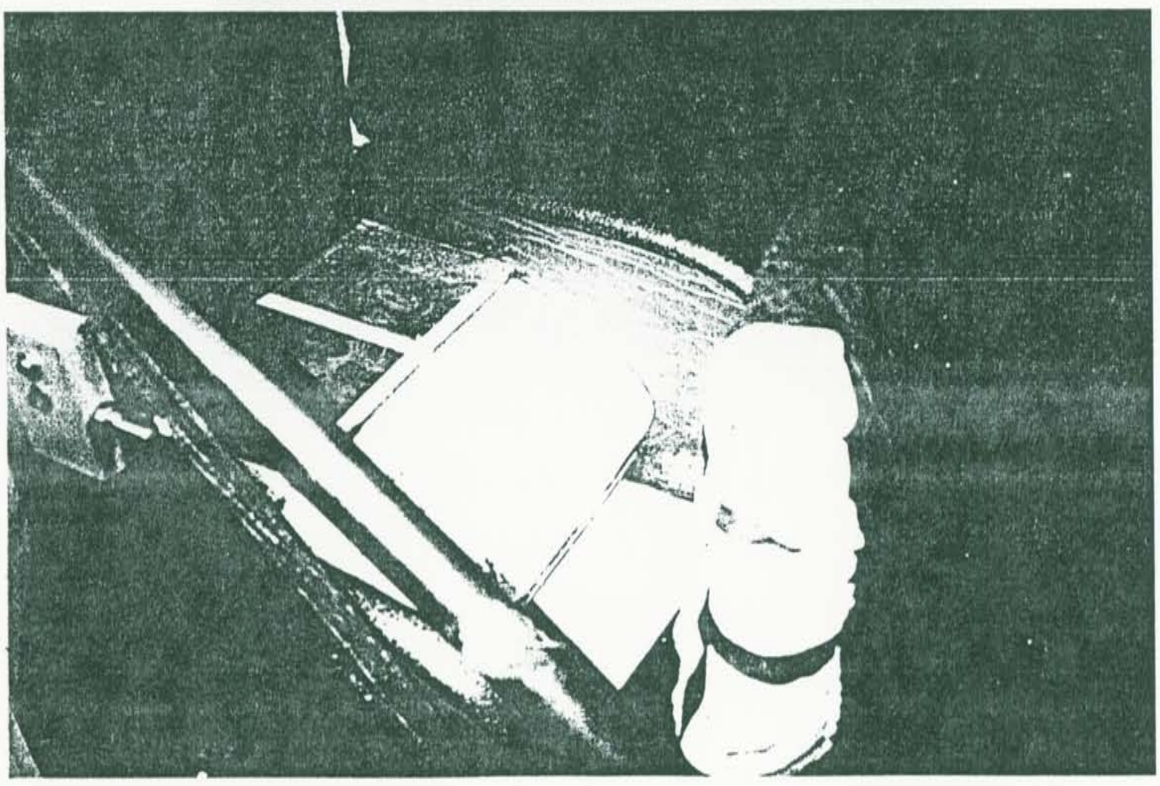


Figure 2.2b - Photographs of the experimental apparatus, showing the handrail and mock-stairway configurations.

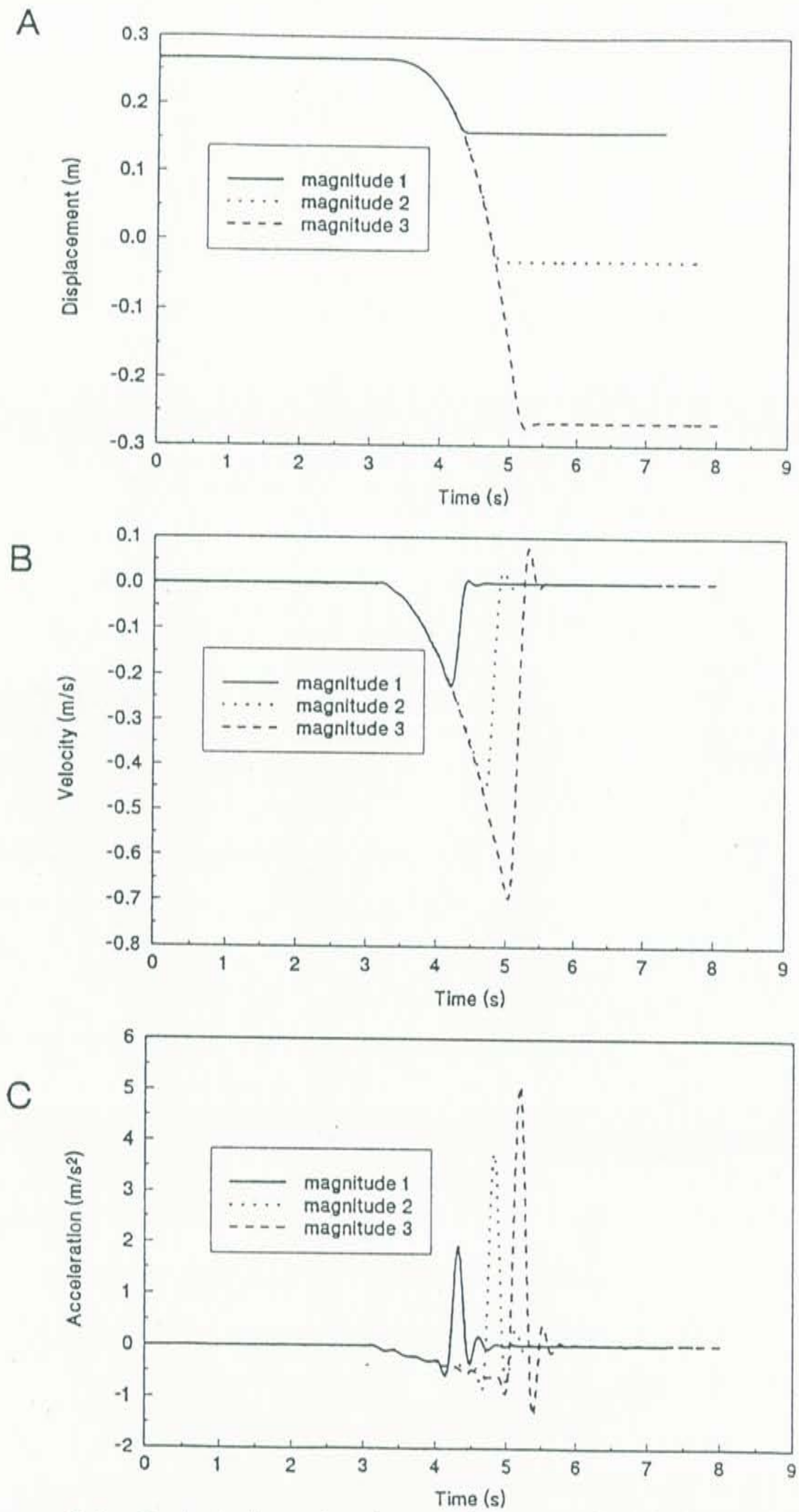


Figure 2.3 - Perturbation waveforms: data recorded during experimental trials, showing the platform displacement (A), velocity (B) and acceleration (C) profiles for the three perturbation magnitudes.

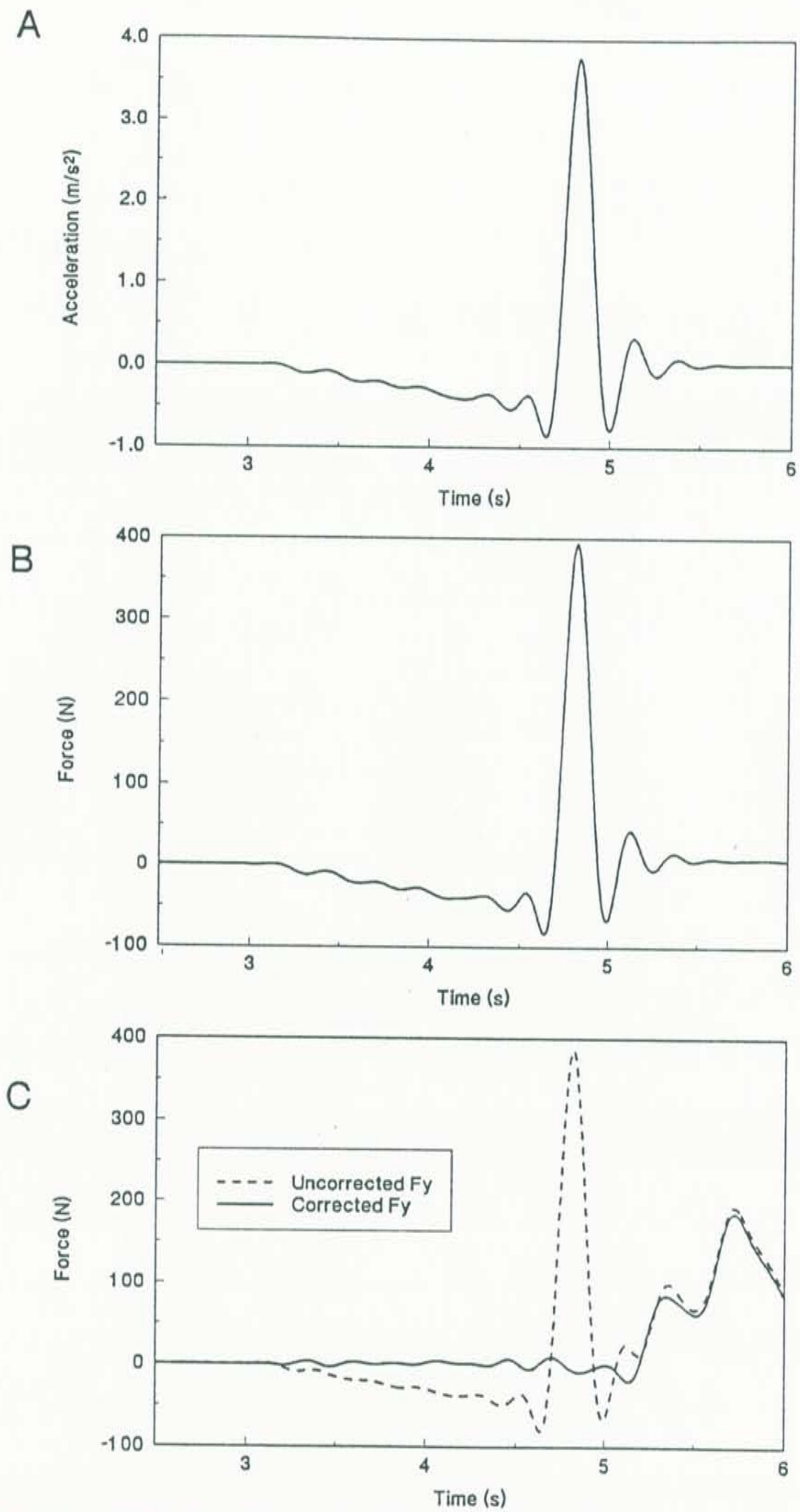


Figure 2.4 - Correction of inertial artifacts: A shows the platform acceleration; B shows the associated artifact in the a-p force (Fy), recorded in the absence of any handrail loading; C shows the a-p handrail force recorded during an experimental trial, before and after correction.

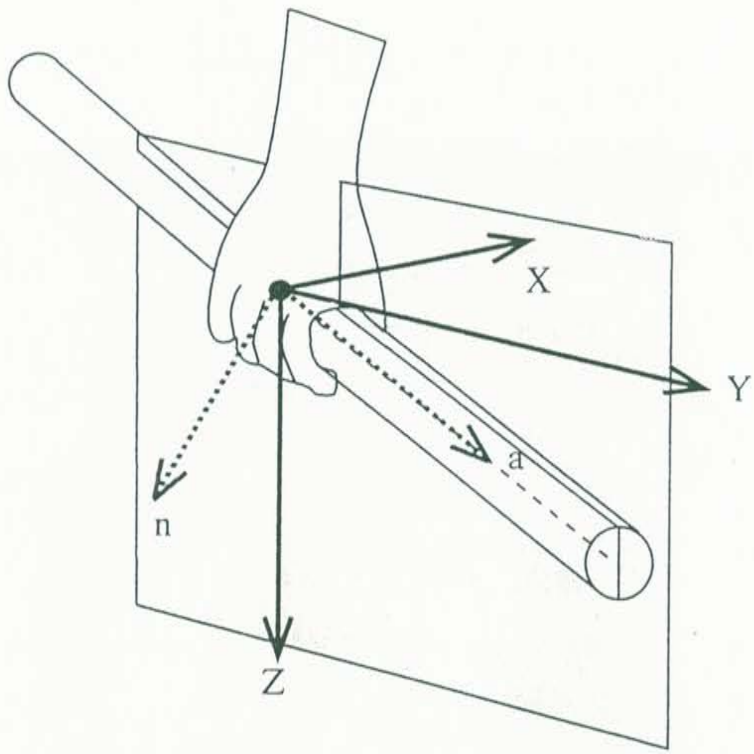


Figure 2.5 - Definition of the handrail force and moment coordinate system. The positive directions correspond to the forces and moments acting *on the handrail*. Note that the sagittal-plane (y-z plane) component is represented using two frames of reference:
1) inertial axes: horizontal (y) and vertical (z), and
2) handrail axes: axial (a) and normal (n).

Table 2.1a: Results of force plate calibration: percentage errors (measured vs actual loads)

VARIABLE		# DATA POINTS	MINIMUM LOAD (N or Nm)	MAXIMUM LOAD (N or Nm)	PERCENTAGE ERROR [100% x (MEASURED - ACTUAL) / ACTUAL]			
					MEAN	STD DEV	MINIMUM	MAXIMUM
AMTI 1:	Fx	20	-475	475	-3.4	1.8	-4.9	2.4
	Fy	20	-475	475	-4.6	3.4	-12.7	4.8
	Fz	192	68	920	0.7	2.2	-4.7	10.1
	Mx	160	-276	414	0.6	3.8	-12.3	16.2
	My	144	-184	368	-2.5	1.8	-12.8	2.7
	Mz	32	-100	100	-2.3	1.6	-4.8	1.5
AMTI 2:	Fx	20	-475	475	2.7	2.2	1.1	8.3
	Fy	20	-475	475	1.5	7.6	-17.1	20.5
	Fz	168	68	920	2.5	1.8	-4.6	7.8
	Mx	112	-184	184	1.7	2.9	-10.2	9.8
	My	144	-552	552	1.3	1	-3.5	6.9
	Mz	26	-154	154	1.6	5.3	-10.9	10.2

NOTE: separate calibration tests were performed for the two force plates on which the handrail was mounted (AMTI 1 and AMTI 2); axes x, y and z refer to medial-lateral (positive to subject's right), anterior-posterior (positive forward) and vertical (positive downward) directions, respectively; applied forces (Fx, Fy, Fz) are listed in N (4.45 N/lb), applied moments (Mx, My, Mz) are listed in Nm (1.36 Nm/ft-lb).

Table 2.1b: Results of force plate calibration: regression of measured vs actual load

VARIABLE	# DATA POINTS	MINIMUM LOAD (N or Nm)	MAXIMUM LOAD (N or Nm)	Y–INTER-CEPT (N or Nm)	SLOPE	R^2
AMTI 1: Fx	20	-475	475	0.091	0.960	1.000
Fy	20	-475	475	0.712	0.961	1.000
Fz	192	68	920	3.096	0.994	1.000
Mx	160	-276	414	2.532	0.998	1.000
My	144	-184	368	-0.360	0.979	1.000
Mz	32	-100	100	-0.163	0.974	1.000
AMTI 2: Fx	20	-475	475	0.542	1.016	1.000
Fy	20	-475	475	-2.897	1.012	1.000
Fz	168	68	920	-0.522	1.027	1.000
Mx	112	-184	184	1.556	1.017	1.000
My	144	-552	552	0.025	1.015	1.000
Mz	26	-154	154	-1.849	1.024	1.000

NOTE: separate calibration tests were performed for the two force plates on which the handrail was mounted (AMTI 1 and AMTI 2); axes x, y and z refer to medial-lateral (positive to subject's right), anterior-posterior (positive forward) and vertical (positive downward) directions, respectively; applied forces (Fx, Fy, Fz) are listed in N (4.45 N/lb), applied moments (Mx, My, Mz) are listed in Nm (1.36 Nm/ft-lb); the slope of the regression is dimensionless; R^2 is the regression coefficient of determination.

Table 2.2: Characteristics of the platform perturbations

VARIABLE	PERTURBATION MAGNITUDE	# DATA POINTS	MEAN	STD DEV	MINIMUM	MAXIMUM
maximum velocity (m/s)	1	64	0.225	0.001	0.223	0.226
	2	64	0.460	0.002	0.456	0.468
	3	64	0.699	0.002	0.694	0.705
maximum deceleration (m/s^2)	1	64	1.905	0.019	1.861	1.961
	2	64	3.725	0.051	3.622	3.821
	3	64	5.253	0.058	5.147	5.524
maximum deceleration jerk (m/s^3)	1	64	23.527	0.338	22.400	24.500
	2	64	42.986	1.222	40.500	45.100
	3	64	53.806	1.391	51.500	58.000
duration of deceleration pulse (s)	1	64	0.221	0.002	0.215	0.225
	2	64	0.236	0.003	0.230	0.245
	3	64	0.262	0.003	0.250	0.265
maximum acceleration (m/s^2)	1	64	0.563	0.014	0.521	0.586
	2	64	0.849	0.052	0.748	0.941
	3	64	0.987	0.054	0.844	1.126
average acceleration jerk (m/s^3)	1	64	0.466	0.011	0.432	0.484
	2	64	0.498	0.031	0.439	0.552
	3	64	0.481	0.026	0.412	0.548
duration of acceleration ramp (s)	1	64	1.209	0.002	1.205	1.210
	2	64	1.704	0.003	1.700	1.710
	3	64	2.051	0.002	2.045	2.055
displacement (m)	1	64	0.108	0.000	0.107	0.108
	2	64	0.298	0.000	0.298	0.298
	3	64	0.534	0.000	0.534	0.535

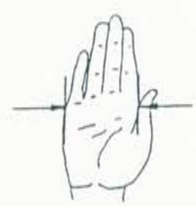
NOTE: The descriptive statistics were calculated using the data collected during the actual handrail experiments, in the 4 subjects (16 trials per subject at each perturbation magnitude).

Table 2.3: Characteristics of the experimental subjects

PARAMETER	SUBJECT 1	SUBJECT 2	SUBJECT 3	SUBJECT 4	MEAN	STD DEV
age (years)	23	26	37	20	27	7
weight (kg)	82	58	78	90	77	14
height (mm)	1840	1740	1774	1690	1761	63
armspan (mm)	1860	1730	1794	1700	1771	71
hand length (mm)	190	170	196	165	180	15
hand width (mm)	115	82	100	95	98	14
finger length (mm)	86	80	89	80	84	5
thumb length (mm)	65	72	72	65	69	4
grip strength (N)	410	430	570	920	583	236

NOTE: The anthropometric parameters are defined below. Grip strength was measured using a Jamar dynamometer (the median score, of three trials, is reported here).

HAND LENGTH



HAND WIDTH

THUMB LENGTH

FINGER LENGTH

3. RESULTS

3.1 Patterns of response

Data pertaining to the frequency of occurrence of the different patterns of response are summarized in Table 3.1. For purposes of analysis, responses were considered to be "valid" provided that the subject grasped the handrail, either without stepping or in combination with a step onto the bottom tread of the mock stairway (see Figure 2.1). The response was considered to be "invalid" if the subject performed a maneuver that was not consistent with an unexpected overstepping of the tread during stair descent. The following specific responses were deemed to be invalid:

1) *stepping with the "wrong" foot*: these responses occurred most commonly when the subject placed the extended foot on the tread cover, and then stepped with the other leg; in a few instances, the subject actually retracted the extended leg, placed it on the top stair tread and then swung the other leg;

2) *overstepping the bottom tread*: the subject stepped with the appropriate leg but missed the bottom tread completely, either by stepping onto the top of the foam rubber "crash pad" (which required the foot to be moved over or through the foam rubber barrier that was placed on top of the crash pad) or by embedding the foot into the end of the crash pad; the strategy of stepping onto the top of the crash pad was seen primarily in a single subject (subject #1);

3) *pushing against the tread cover*: the subject was able to balance without stepping but was observed to push against the tread cover with the extended foot, either by plantarflexing the foot or by stepping onto the tread cover; this strategy was seen predominately in two subjects (#2 and #4);

4) *grabbing the platform walls*: the subject grabbed onto the edges of the protective walls and/or railings that surround the moving platform, using the left hand; this response was very uncommon, occurring in only two trials;

5) *other invalid response:* this category included trials where the subject appeared to grab the handrail unnecessarily (i.e. as an "afterthought"), after the body motion had already been arrested (this occurred only at the smallest perturbation); also, trials where the subject stepped, or hopped, sideways on the top stair tread (these occurred only in the "far" stance position, and most commonly when the feet were obstructed);

6) *early arm EMG:* reactions to unpredictable perturbations require at least 40ms (for neural processing and signal conduction) before the muscles are activated; therefore, trials showing muscle activation (EMG) occurring any earlier than this, relative to the onset of the platform deceleration, were considered to represent an attempt to respond in an anticipatory, or predictive, manner; anticipatory activity was observed in 27 (14%) of 192 trials, and the great majority of these (n=21) occurred at the largest perturbation; the percentages indicated in Table 3.1 include only those trials that were not already excluded for the reasons listed above.

Over all three task conditions (main experiment, feet-obstructed trials and hand-on-rail trials), approximately 60% of responses were valid, with about half of these responses involving stepping. The vast majority of the non-stepping responses occurred at the smallest perturbation. There were no trials where the subject was unable to establish a grip on the handrail; however, there were two trials where the initial hand trajectory missed the handrail, and there were four trials where the hand was moved after initial contact occurred. In all of these cases, the error in the response was corrected and a functional grip was eventually established. There was only one trial, out of a total of 192, where the subject was unable to maintain a grip after it was established (i.e. the hand "broke free").

In the main experiment, subjects were instructed to try to keep their balance by grabbing the handrail and to try to avoid stepping if possible. As indicated in Table 3.1, the subjects were successful in avoiding stepping (without resorting to "invalid" maneuvers) in about half of the small-perturbation trials. In contrast, they avoided stepping in only 8% of the medium-perturbation trials, and were never able to accomplish this at the largest perturbation. In a relatively small percentage of trials (22/144 or 15%), the subjects grabbed the handrail with the left hand subsequent to grabbing with the right hand. Invariably, the left hand contacted the handrail well after the initial contact with the right hand: the interval between right and left contact ranged from 0.47 to 1.7 seconds (mean 0.93s, SD 0.36s).

In the feet-obstructed trials, further measures were taken to discourage stepping: a small foam-rubber barrier was placed immediately in front of the feet, and subjects were allowed to stand with both feet on the uppermost stair tread prior to the perturbation. Overall, this approach was successful in eliciting "valid" no-step responses in 15 of the 24 trials (63%), with a 100% success rate at the small perturbations. In cases where subjects did not step, the net effect of grabbing the handrail with the right hand, in combination with the momentum imparted to the center of mass by the perturbation, was to cause the body to swing around laterally toward the handrail. In the absence of stepping, the subject was thus forced to use the left hand to either grab the handrail or to push on the foam crash pad, in order to arrest the angular momentum of the body. The subject contacted the handrail with the left hand in 17 of 24 trials. The delay between right and left contact was very similar to that observed in the main experiment, as described above.

In the hand-on-rail trials, subjects were allowed to grasp the handrail prior to onset of the perturbation. In these trials, they were able to recover equilibrium without stepping, or resorting to "invalid" maneuvers, in 42% of trials, in comparison to the 20% success rate that occurred during the main experiment. This difference in frequency of grasp-only trials was statistically significant (chi-square test; observed level of significance, $p=0.02$). Subjects did not contact the handrail, or crash pad, with their left hand in any of the hand-on-rail trials.

To determine whether the subjects would have been able to recover balance simply by stepping, without any use of the arms, each subject performed six trials in the absence of a handrail. In general, the subjects were not able to recover equilibrium simply by stepping. In 46% of these trials (11/24), the subjects used an "invalid" maneuver, either grabbing the walls of the moving platform with the left hand (n=8) and or pushing against the tread cover with the extended foot, in order to prevent stepping (n=3, all at the small perturbation). Moreover, in 29% of the trials (7/24), the subjects fell into the crash pad, contacting the pad with the arms (n=3) or with the body (n=4). In contrast, subjects fell into the crash pad in only 6% (9/144) of the main-experiment trials, and none of these trials involved a complete loss of equilibrium, i.e. contact with the body. If contact with the crash pad is considered to represent a "fall", then subjects fell in 54% (7/13) of valid-response trials when the handrail was absent, and in only 7.5% (6/80) of valid-response trials when the handrail was present (main experiment). This decrease in frequency of "falling", due to the grabbing of the handrail, was highly statistically significant (chi-square test: $p<0.001$).

3.2 Patterns of force and moment generation

Example data, showing handrail forces and moments, as well as the motion of selected body markers, are displayed in Figures 3.1 to 3.4. Figures 3.1 and 3.2 show representative main-experiment trials where the subject responded by grasping the handrail only (Figure 3.1), or by grasping and stepping (Figure 3.2) Figure 3.3 shows a feet-obstructed no-step trial in which the subject grasped the handrail with the left hand subsequent to the initial grasping with the right hand. Figure 3.4 shows a hand-on-rail trial where neither stepping nor left-hand contact occurred.

Descriptive statistics summarizing the handrail-force data are provided in Tables 3.2 to 3.4, for the main-experiment, feet-obstructed and hand-on-rail trials, respectively. Each table comprises three sections, describing: a) peak absolute force, b) peak relative force (normalized with respect to body weight), and c) timing of the peak force. Descriptive statistics summarizing the kinematic and electromyographic (EMG) data are provided in Tables 3.5 to 3.7, for the main-experiment, feet-obstructed and hand-on-rail trials, respectively. It should be noted that, in this report, any discussion of the direction of the forces and moments refers to the direction of the forces and moments exerted by the hand on the handrail, unless noted otherwise (see Figure 2.5).

This report focuses on the handrail forces, rather than the pure moments generated at the hand. In general, the pure moments tended to be relatively small, in terms of their expected stabilizing effect on the body. If one approximates the dynamics of the falling body as an inverted pendulum rotating about the stance foot, it is the moment about the foot axis that acts to counter the falling motion of the body. Peak values of pure hand moment, on average, ranged from approximately 5 to 35Nm, depending on the axis of rotation and the perturbation magnitude. By way of comparison, the moments about the foot axis generated by the handrail reaction forces would tend to be an order of magnitude larger (e.g. a 200 N force, acting at a moment arm of 1m, would generate a moment of 200Nm about the foot axis).

Nonetheless, it can be noted that the largest pure moments tended to occur about the positive m-l (x) axis, the negative a-p (y) axis and the positive vertical (z) axis: the corresponding mean pure moments were 26Nm (SD 11), 21Nm (SD 12) and 13Nm (SD 9), respectively, at the medium perturbation magnitude (main experiment). Values at the large perturbation were about 50% higher, and occasionally ranged as high as 60Nm in individual trials; however, such large pure moments were relatively rare. Axial moments about the longitudinal axis of the handrail, which might be of relevance to the problem of hand slippage, tended to be modest, with means in the large-perturbation trials (main experiment) of only 12Nm (SD 6) and 8Nm (SD 3), for the negative and positive axis directions, respectively (the corresponding maximum values, from individual trials, were 22 and 14Nm). As noted in the Methods (Section 2.8), methodological problems limit the accuracy of the pure-moment measurements, and errors as large as 5 to 10 Nm would not be unexpected; therefore, the above values should be viewed as approximate estimates.

3.2.1 Main experiment

The most consistent feature of the force generation seen in the main experiment was the tendency for the axial handrail force component to be directed forward along the handrail throughout the duration of each trial. The data in Table 3.2a provide supporting evidence for this observation. In over 80% (67/80) of the valid-response trials, there was negligible force (peak force<10N) in the backward axial direction, and the average magnitude of the backward peak, in the remaining trials, was only about 10% of the forward peak (16N versus 136N).

As is evident in Figures 3.1 and 3.2, the hand tended to move very quickly to the handrail following onset of platform deceleration. Thus, the hand was anchored very early, while the body continued to fall forward. This caused the hand to be situated posterior to the shoulder and trunk by the time the peak force was generated; therefore, the generation of the axial force would appear to involve a pulling, rather than pushing, action (see Figures 3.1 and 3.2, panel C),

The normal force component tended to show peaks of similar magnitude in both positive and negative directions. Typically, the first peak in the response was positive, indicating that the hand was pushing down against the surface of the handrail. This was usually followed by a phase of negative force, during which the hand pulled up and away from the handrail.

The lateral force component was also quite consistent, almost always being directed to the subject's left, i.e. the hand pulling away from the handrail. This is likely a consequence of the fact that the forward momentum of the body tends to cause the body center of mass to rotate, in the transverse plane, about the anchor point provided by the handrail grip. The lateral force generated at the hand provides the centripetal reaction necessary to sustain the rotation.

When the sagittal-plane force is decomposed into a-p and vertical components (rather than axial and normal), it is seen that the a-p force tended to be directed primarily in the forward direction, and that the vertical force tended to be directed downward. Thus, the reaction forces that acted on the hand were directed so as to counter the forward and downward motion of the center of mass. These generalizations are supported by the data in Table 3.2a, which show that the mean forward force was about six times as large as the mean backward force (138N versus 21N). Peak upward vertical forces were very often negligible (<10N in 52 of 80 valid-response trials) and, in trials where significant upward force did occur, the magnitude tended to be only half as large as the downward force (mean of 57N versus 118N). In terms of timing, the first peak tended to occur in the vertical force, with the peaks in the a-p and m-l forces usually occurring somewhat later. The peak vertical, a-p and m-l forces tended to be similar in magnitude, and hence tended to contribute equally in determining the resultant force. The resultant force often showed two distinct peaks, corresponding approximately to the peaks in the vertical and the a-p/m-l force components (e.g. see Figure 3.2A).

3.2.2 Feet-obstructed trials

The pattern of force generation recorded in the feet-obstructed trials was similar, in many respects, to that seen in the main-experiment trials, particularly in terms of the m-l, a-p and axial components; however, there did appear to be some differences in terms of the vertical and normal force components. Although one must be cautious in attempting to generalize from such a small number of trials, it did appear that the subjects were more likely to pull vertically upward in later stages of the response (as noted earlier, upward vertical forces tended to be small, or non-existent, in the main-experiment trials). In terms of the normal component, there appeared to be a tendency for the negative peak (pulling up on the handrail) to be much larger in magnitude than the positive peak (pushing down on the handrail). These differences in pattern of force generation would appear to correspond to more exaggerated efforts to prevent stepping by counterbalancing the moment about the feet due to the weight of the forward-falling body. Because the hand is situated posterior to the "fulcrum" provided by the stance foot, the hand must pull upward on the handrail.

3.2.3 Hand-on-rail trials

For the hand-on-rail trials, the pattern of force generation, in terms of the m-l, a-p and axial components, was similar to that observed in the main-experiment and feet-obstructed trials. Once again, the main differences appeared in the vertical and normal force components. In these trials, the initial peak in the vertical and normal forces tended to be negative, indicating that the hand was tending to pull up on the handrail from the onset.

3.3 Analysis of contributing factors

Statistical analyses of the factors that could potentially affect the use of the handrail were restricted primarily to the main-experiment trials, because of the limited numbers of trials performed during the other task conditions. The following force variables were analyzed: 1) peak leftward m-l force (positive Fx), 2) peak forward a-p force (positive Fy), 3) peak downward vertical force (positive Fz), 4) peak forward axial force (positive Fa), 5) peak upward normal force (negative Fn), 6) peak downward normal force (positive Fn), and 7) peak resultant force. Note that the m-l, a-p, vertical and axial forces are represented only by the peaks in one direction, because the forces generated in the opposite direction were often negligible and, when present, tended to be much smaller in magnitude (see Table 3.2). In addition to the force variables, the time required for the hand to contact and grasp the handrail and the location of the hand (i.e. the wrist marker) at the time of full grasp (a-p and m-l coordinates) were also analyzed. For each of the dependent variables listed above, a four-way repeated-measures ANOVA was performed, to determine main effects due to perturbation magnitude, stance position, stance leg and "round" (i.e. repeated trial) number. Results were considered to be statistically significant at $p<0.05$.

3.3.1 Influence of perturbation magnitude

Perturbation magnitude had a statistically significant influence on all of the force-magnitude variables. As can be seen in Tables 3.2a and 3.2b, there was a consistent tendency for the magnitude of the peak force to increase with perturbation magnitude. The most pronounced effects were seen in the a-p and axial forces (p's<0.01). Each increment in perturbation level corresponds to an equal change (approximately 0.25m/s) in maximum platform velocity, and in most cases, the effect on the mean normalized peak force appeared to scale linearly with respect to the platform velocity. The notable exception was the m-l force, which showed a sizeable difference between small (magnitude 1) and medium (magnitude 2) perturbations, but very little difference between the medium and large perturbations. As discussed in Section 3.3.4, this finding may be related to the influence on the m-l force due to stepping, which was much more likely to occur at the larger perturbations.

Perturbation magnitude also had a statistically significant influence on the timing of the grasp and on the a-p location of the grip. As might be expected, there was no effect on the m-l grip coordinate. As can be seen in Table 3.5, the hand tended to grasp the handrail about 3cm farther forward along the rail, on average, with each increment in perturbation magnitude ($p<0.001$). At the same time, the effect of each increment in perturbation magnitude was to decrease the time required to contact the handrail by 30 to 70ms ($p<0.01$). Completion of the grasp typically required an additional 120-130ms, independent of the perturbation magnitude.

3.3.2 Influence of initial stance conditions

The initial stance position, i.e. the proximity of the subject to the handrail, had a statistically significant effect on only two of the force variables: m-l force and a-p force. Interestingly, the leftward m-l force (positive Fx) tended to be 50% larger, on average, when the subject stood farther away from the handrail ($p<0.01$). The average normalized m-l forces were 20% and 13% of body weight, for the "far" and "near" positions, respectively. Conversely, the forward a-p force (positive Fy) tended to be slightly smaller when the subject stood in the "far" position ($p<0.05$). The average normalized forces were 16% and 19% of body weight, for the "far" and "near" positions, respectively.

The position-related effect on m-l force may be related, once again, to the tendency of the center of mass to rotate, in the transverse plane, about the "anchor point" established by the gripping of the handrail. When the body is displaced laterally, the angular momentum of the center of mass, as it pivots about the anchor point is increased; hence, larger centripetal reaction forces (largely m-l) will be exerted on the handrail. It is not clear, however, why the a-p force would tend to decrease in this situation. It should be cautioned that the size of this latter effect was small, and could represent a "false positive" result.

Not surprisingly, the time required to contact the handrail increased when subjects stood farther away from the handrail ($p<0.01$). The average values, for the "near" and "far" positions were 0.53s and 0.67s, respectively. In addition, there was a significant effect due to initial stance position on the m-l location of the grip, as indicated by the wrist marker coordinates ($p<0.01$). In trials where the subject stood farther away, the hand tended to move in a relatively straight trajectory toward the handrail, and initial contact occurred with the extended thumb "hooking" onto the inside edge of the rail and/or fingers wrapping around the outside of the rail. Thus, the m-l position tended to be to the left of the rail center-line (mean m-l position, -0.03m). Conversely, when the subject was standing close to the handrail, the hand tended to move in a more curved trajectory, up and over the rail, so that the "angle of attack" was more nearly vertical, with initial contact occurring in the "notch" between the thumb and the index finger. In this situation, the m-l position tended to be to the right of the handrail center-line (mean m-l position, +0.03m).

There were no statistically significant effects due to the initial stance leg (right or left) on any of the force variables that were analyzed. Timing of the grasp was also unaffected. Interestingly, there was one small, but statistically significant, effect on the grip location ($p<0.01$). In trials where the subject stood on the right leg, the hand tended to be situated to the right of the handrail center-line (mean m-l position = +0.004m). The opposite was true when the subject stood on the left leg (mean m-l position = -0.003m). The explanation for this finding is not obvious; however, once again it should be cautioned that the size of this effect was small, and could represent a "false positive" result.

3.3.3 Trial-to-trial variation

The statistical analyses showed no evidence of systematic changes, due to the repeating of the trials, in any of the force, grip location or timing variables that were analyzed. If such effects were present, they were likely masked by the relatively large trial-to-trial variability that was observed.

Because of the relatively high frequency of "invalid" responses, there were insufficient numbers of trials to calculate reliable estimates of trial-to-trial variability, within each individual subject, for each specific set of test conditions. Instead, within-subject estimates of variability were constructed for each perturbation magnitude by pooling all of the available data (2 stance positions x 2 stance legs x 3 rounds). The variability was quantified in terms of the coefficient of variation, or COV (COV = standard deviation / mean). For the force variables, the COVs tended to be moderate, ranging from approximately 20% to 40% at the medium and large perturbation magnitudes. The variability tended to be somewhat higher at the small perturbation, however, with COVs ranging as high as 80%. The variability in the timing of handrail contact tended to be less than that observed in the force variables, with COVs ranging from about 10% to 20% (COVs were not calculated for the grip location, because the mean level is defined relative to an arbitrary reference point). Example data, illustrating the trial-to-trial variability in the force variable are provided in Figure 3.5.

In general, variability in the force, grip location and timing variables tended to be quite high when measured across all subjects. An indication of this variability is provided by the standard deviation values listed in Tables 3.2 to 3.7. At each perturbation magnitude, across-subject COVs were typically on the order of 40 to 60%.

3.3.4 Influence of stepping

About 60% of the "valid" main-experiment trials included both stepping and grasping reactions. To examine whether stepping had an influence on the measured handrail force and grip variables, further repeated-measures ANOVAs were performed; stepping (versus not stepping) and perturbation magnitude were included as the factors in these analyses. Because there were no "valid" non-stepping responses in the large-perturbation trials, the analyses were limited to the small and medium perturbations. Interestingly, at each perturbation magnitude, there was a tendency for most of the force components to show larger peaks when stepping occurred. In fact, the only force components that did not show statistical evidence of an association with stepping were the axial and a-p forces. The most pronounced effects, due to stepping, were seen in the downward normal component and the resultant force (p's<0.01). For the small perturbations, the mean resultant force increased from 16% of body weight to 23% as a result of stepping; for the medium perturbations, the corresponding values were 23% and 33%. There were no statistically significant differences, due to stepping, in the grip location; however, the initial contact tended to occur more rapidly in the trials where the subjects did not step ($p<0.01$). The average delays in contact time, associated with the stepping responses, were 40ms and 100ms, for the small and medium perturbations, respectively. The corresponding delays in time required to achieve a full grip were even larger: 90ms and 150ms.

One might have predicted that subjects would step in trials where they failed to generate sufficient stabilizing handrail force. The present findings actually showed evidence of the opposite trend, i.e. larger forces in stepping trials; however, it is important to note that the present analyses looked only at the peaks in the force generation, which tended to occur rather late, in comparison to the time at which the "decision" to step may have been made (our previous studies of stepping suggest that this may occur as early as 150ms after onset of perturbation[23]). The present results could indicate that the "decision" to step was necessary because of a delay in the grasping response; however, it is also possible that an early "decision" to step reduced the urgency of the need to grasp the handrail and hence the arm may have been moved less rapidly. Further analyses of the arm movement and the muscle activation, in both legs and arms, may help to resolve these questions. In addition, it may be useful to examine the early changes in handrail force and, in particular, the time-integral of the force (i.e. the impulse), which defines the change in center-of-mass momentum that results from the force generation. These analyses are beyond the scope of the present report. One conclusion that can be drawn from the present analyses is that at least some of the peak force components (m-l, normal and vertical) are influenced by the biomechanical demands of the stepping response. Conversely, the a-p and axial force components appear to be relatively independent of whether or not a step was executed in conjunction with the grasping response.

To determine whether the results of the first set of analyses, regarding the influence of perturbation magnitude and initial stance conditions, were confounded by the presence or absence of stepping, these analyses were repeated while including only the stepping trials (corresponding analyses could not be performed for the non-stepping trials, because of the limited number of such trials). Not surprisingly, in view of the effects of stepping outlined above, the primary changes occurred in the analyses of the m-l, normal and vertical force components. For the normal and vertical force components, the exclusion of non-stepping responses tended to reduce the statistical significance of the effect due to perturbation magnitude; however, there was still a trend for the forces to increase with the size of the perturbation. In contrast, for the m-l force, there was no longer any evidence to support a perturbation-magnitude effect ($p=0.2$). In addition, there was no longer any evidence to support an effect due to stance position ($p=0.4$). Earlier, it was suggested that the effect due to stance position seen in the initial analyses was related to the tendency of the body to rotate about the anchor point created by the handrail grip. The present findings would suggest that the act of taking a step tends to counters this tendency, and that the m-l handrail force is then defined primarily by the biomechanical demands of the stepping response (e.g. the need to control lateral stability during the swing phase) rather than the magnitude of the perturbation.

3.3.5 Influence of task constraints

In the final set of analyses, the three different task conditions--main experiment, feet-obstructed trials and hand-on-rail trials--were compared by means of ANOVA. Because of the small numbers of trials performed in the latter two tasks, it was necessary, for each perturbation magnitude, to pool the data pertaining to the different stance conditions. As a result, however, caution should be exercised in interpreting the results of the analysis, which may be confounded by the failure to account for potential effects due to stance condition. Furthermore, it should be noted that failure to show significant findings, in some cases, could be due to the limited statistical power that the small sample provides.

In comparing the tasks, we anticipated that the foot obstruction, in preventing stepping, would require larger forces to be generated in order to recover equilibrium. Conversely, we expected to see lower forces in the hand-on-rail trials, because, in this condition, the hand has the potential to begin generating stabilizing force much earlier, and might thereby act to arrest the body motion before it starts to pick up momentum during the forward fall.

In general, the force data presented in Tables 3.2 to 3.5 would suggest that there were actually few differences between the main-experiment and feet-obstructed trials. On the other hand, a number of the force components generated during the hand-on-rail trials appeared to be substantially lower, when compared to the other two tasks. The statistical analyses provided evidence to support the latter observation, but only for the m-l force component. The mean normalized m-l force was approximately twice as large in the main-experiment and feet-obstructed trials, in comparison to the hand-on-rail trials (14-16% of body weight versus 8%). The upward normal force component actually showed the opposite trend: it was significantly lower in the main experiment (12% of body weight, versus 21% in both the hand-on-rail and feet-obstructed trials).

The task also had a statistically significant effect on the a-p grip location: on average, the grip was placed farther forward during the main experiment and the hand-on-rail trials, in comparison to the feet-obstructed trials (means of 0.283m and 0.269m, versus 0.212m; $p<0.01$). There were no significant differences in the m-l grip location, between any of the tasks, and the timing of initial handrail contact did not differ significantly, in comparing the main-experiment and feet-obstructed trials.

The differences in response observed during the feet-obstructed trials appear to be consistent with a more exaggerated effort to prevent stepping. As discussed in Section 3.2.2, the moment about the feet generated by the handrail force acts to counter the moment due to the forward displacement of the body weight. In the feet-obstructed trials, where subjects are most highly motivated to avoid stepping, they seem to maximize the stabilizing moment in two ways: by increasing the upward normal force on the handrail, and by increasing the moment arm with respect to the feet, i.e. by locating the grip in a more posterior location.

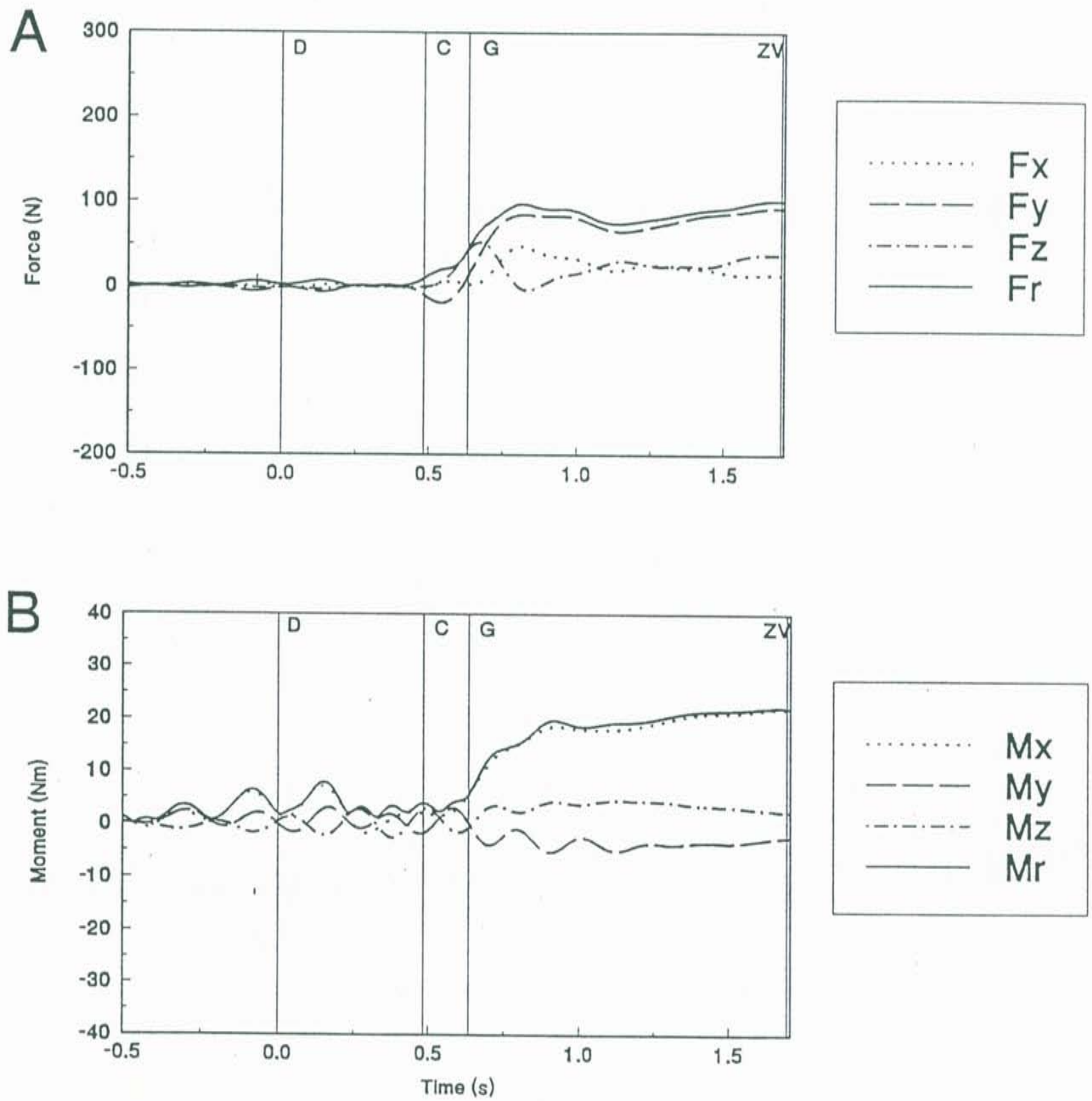


Figure 3.1 - Data from a representative main-experiment trial: grasp-only (no-step) response; task conditions - small perturbation, "close" position, left stance leg; see the explanatory note following Figure 3.4 for figure details.

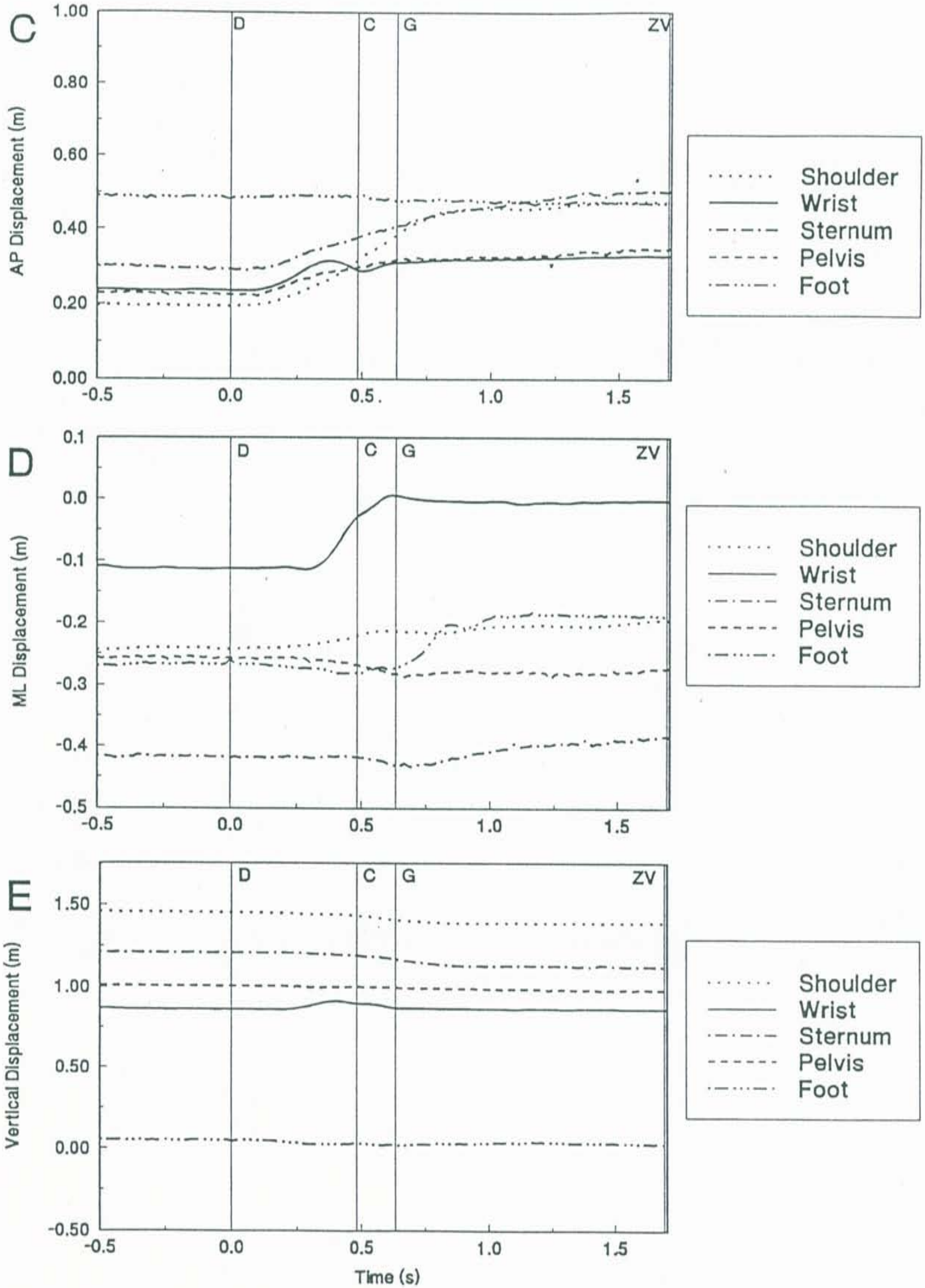


Figure 3.1 continued

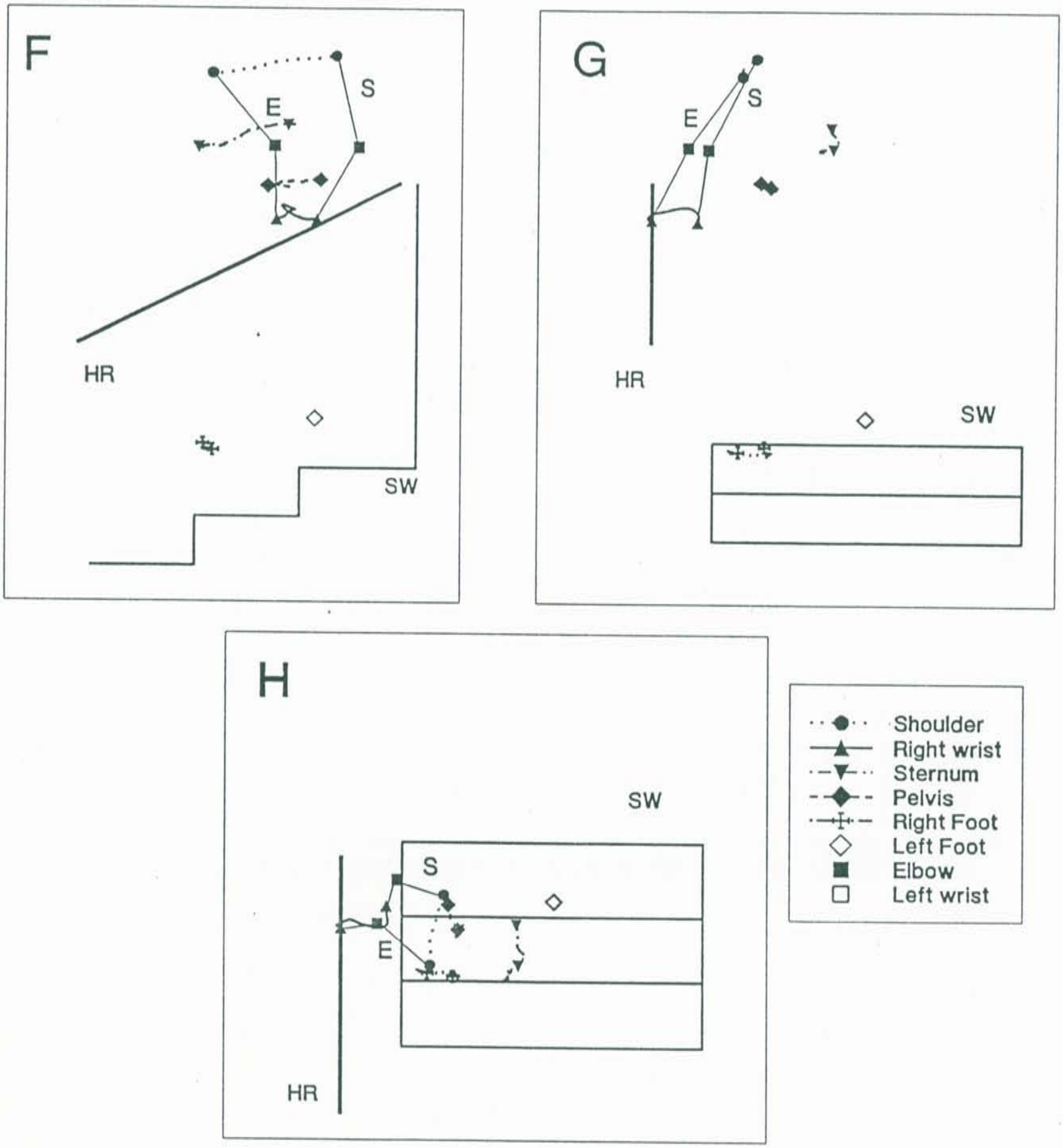


Figure 3.1 continued