

Biomechanics of Balance Recovery by Handrail Grasping

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ABSTRACT

While public transportation continues to grow in relevance, few ergonomic and biomechanical studies have examined passenger safety, especially related to falls from standing, and the role of handrail grasping in balance maintenance. In this thesis, I examined forces from upper limbs (from a load cell), lower limbs (from a force-plate) and muscle responses (from EMG) involved in balance recovery during simulated vehicle acceleration and deceleration pulses. Subjects stood (forward or sideways) on a platform holding a handrail in two different positions (shoulder height and overhead). A linear motor accelerated the platform at high and low perturbation magnitudes. Hand configuration, perturbation direction and perturbation magnitude influenced hand forces, COP displacement and muscle activation. Forward stance while holding the handrail at shoulder height increased the musculoskeletal demands of balance maintenance. Improved knowledge from this project should help guide the design of safer vehicles.

“Although no one can go back and have a new beginning, anyone can start today
to make a new ending”

(Chico Xavier)

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CHAPTER 1: INTRODUCTION

1.1 Injuries related with public transportation usage

Public transportation, such as buses, subways and trains, is an important component in planning and managing urban regions (Golob and Hensher, 2007, Murray et al., 1998). Approximately ten percent of the Canadian population used public transportation between 2006 and 2007, and over the same one-year period, there was an increase of 4.9% in its use (Statistics Canada, 2007). The growing urban population, added with traffic congestion and the need to reduce energy consumption and environmental pollution, motivates increased reliance and growth in the use of public transportation (Statistics Canada, 2008).

Unfortunately, there is a high frequency of injury related to use of public transportation (Helpern et al, 2005; Wahlberg, 2004), which can be divided into two major categories: injuries due to crash incidents and non-collision injuries. Injuries due to crash incidents are those where the passenger is injured during vehicle collision or when a vehicle hits a pedestrian. Non-collision injuries are those where the passenger is hurt during normal movements of the vehicle. This can happen, for example, when the passenger is boarding or unboarding the vehicle, walking or simply standing inside the vehicle, and these injuries are typically from falling or hitting objects inside the vehicle. The expected increase in public transportation usage reinforces the need to identify and address safety issues related to both classes of injuries for the design of safer vehicles.

The causes and consequences of non-collision injuries in buses in the city of Uppsala, Sweden, were explored by Wahlberg (2004), who conducted a four year study collecting reports from the city bus company. Of the 129 injury cases reported during this period, 29% occurred at bus stops, probably due to sudden decelerations. Bjornstig et al., (2005), in a ten-year follow-up study, showed that from 284 bus and coach injuries reported during that period, non-collision incidents were responsible for 54% (n = 154) of all cases.

Halpern et al. (2005), aiming to identify the cause of non-collision injuries, distributed questionnaires to six medical centers around the city of Tel Aviv. The authors observed 120 bus injuries over six months. From these, 14.2% of cases were admitted to hospital. The majority of injuries occurred during bus acceleration/deceleration pulses (51.2% of all cases). In particular, 56% occurred in standing passengers, 25% in passengers walking from one place to another, and 19.2% in those sitting. The most common site of injury was the limbs (33.3%), followed closely by head (29%) and spine (22%).

Kirk & Grant studied bus injuries in the UK between the period of 1999 and 2001 (Kirk & Grant, 2001). They showed that, of a total of 9100 injuries occurring per year, 5.6% involved serious injuries, and 64.3% of these occurred in non-collision incidents, and 39.0% when the passenger was standing. The authors suggested a series of improvements to the UK Department for Transport that could lead to fewer injuries including changes to distances between seats, door step height, surface hardness and geometry (sharp corners) and bus driver hours.

1.2 Cause and risk factor for injuries in public transportation

When standing on buses, subways or trains, vehicle acceleration and deceleration induces challenges to postural stability, due to the inertial force applied to body segment which are typically countered by generating external forces between the feet and the ground, or between the hand and a handrail. Palacio et al., (2009) conducted a series of experiments where they measured acceleration peaks during typical bus routes in Dublin. The authors reported average peak of acceleration of 1.4 m/s^2 and 1.6 m/s^2 during the vehicle normal movement, and peak accelerations of 3.2 m/s^2 , during braking. DeGraaf and Weperen (1997) conducted a series of experiments to explore the ability of humans to maintain an upright posture when standing without grasping on a treadmill that applied random accelerations to the feet ranging in magnitude from 0.3 to 1.6 m/s^2 . Trials where subjects needed to take a step, to hold on the treadmill hand support, or be caught to avoid a fall, were regarded as failure to maintain balance. Pulses were applied in four different directions (forward, backward and lateral). The authors found that the threshold to failure was smaller for lateral accelerations (0.33 m/s^2) than forward and backward (0.54 m/s^2 and 0.61 m/s^2 , respectively). In the second set of experiments, the authors measured bus, trams, express tram and metro acceleration profiles in real situations. They found peak accelerations between 1.0 m/s^2 to 2.0 m/s^2 , and jerk magnitudes of 1.5 m/s^3 and 3.5 m/s^3 for longitudinal and transverse directions, respectively. These values are higher than those necessary to cause individuals to lose balance if standing and not grasping a support. Although DeGraaf and Weperen study contributes to the understanding of balance recovery on moving vehicles, the participants were always required to recovery

balance without grasping a handrail support, which does not fully reflect real life situations. In the present work, I built on their study to examine balance maintenance while standing and grasping a handrail during simulated vehicle accelerations.

1.3 Biomechanics of standing balance

1.3.1 Posture and balance during quiet stance

Posture is defined as the different orientations and angles of the body segments relative to the gravitational vector (Winter, 1995), while balance is defined as the ability of the individual to maintain the centre of mass (COM) within the base of support and thereby prevent a fall (Massion, 1992; Pai & Patton, 1997).

To perform any task that require a change in posture, the nervous system collects and interprets information from the somatosensory, vestibular, and visual systems, allowing multi-joint coordination, and environmental adaptation to complete the task successfully (Horak and Kuo, 2000). Indeed, this information is essential even for the control of body sway during quiet standing, which is a commonly relied upon measure of balance control.

The primary variable recorded during laboratory-based tests of sway during quiet stance is the location of the centre of pressure (COP) between the feet and ground. These measures are normally acquired from a force plate (Winter, 1995), and analyzed to determine the variation in COP location in the anterior-posterior (a-p) and the mediolateral (m-l) directions. According to Winter et al., (1996), a change in COP

location represents a net response from the neuromuscular system to control the location of the centre of mass. Furthermore, the COP moves in phase with the COM during quiet standing, suggesting that muscle stiffness alone may control balance during standing (Winter, 1998). However, other authors have argued that processing of sensory information by the central nervous system plays an important role in control of balance during quiet standing, and that the in-phase movement of the COP and COM is simply a result of physics (Morasso and Shieppati, 1999).

1.3.2 Balance maintenance through feet-in-place strategies

One of the first researchers to use a movable platform to induce a whole body perturbation to balance was Nashner (1971) with the goal of developing a better understanding of the role of vestibular function in the control of body sway. His platform could displace horizontally, rotate, and in some cases be displaced vertically. Since then, other authors have also used perturbation platforms to study the organization of movement strategies during perturbed stance (Nashner, 1976; Diener et al., 1982; Allum and Pfaltz, 1985; Horak and Nashner, 1986; Carpenter et al., 2004). Nashner's early work in this area led to the observation that, following a sudden platform translation, the CNS organizes muscle activation by first stabilizing the joint closest to the perturbation, followed by activation of muscles spanning the knee, hip and spine in a bottom-up sequence (Nashner, 1982).

Horak and Nashner (1986) extended these observations by describing two muscle synergies for stabilizing balance following horizontal platform translation: the

ankle strategy and the hip strategy. The ankle strategy involves contraction of the muscles spanning the ankle to bring the centre of mass back to a position of stability. The ankle strategy is often used when the perturbation is small, or the base of support is firm. Muscle activation tends to commence 90-100 ms after the onset of the perturbation, starting on the leg (tibialis anterior or gastrocnemius), and followed by the thigh (hamstrings or quadriceps) and finally the paraspinal or abdominal muscles. If the subject experiences a larger perturbation, or if the support surface is narrow, he/she may adopt a hip strategy for balance maintenance. This strategy is characterized by large and rapid motions of the hip in the same direction as platform translation, which moves centre of mass back to a stable position, inside the base of support. Muscle activation again commenced 90-100 ms after the perturbation onset, but starts in the trunk muscles, followed by thigh muscles. Calf muscle activation is minimal. Also, forward platform translation generates activity in the posterior muscles, while backward translations perturbation generates activity in the anterior muscles. Although perturbation magnitude and surface configuration may influence the adoption of different strategies, individuals often use a mix of ankle and hip strategies.

1.3.3 Change in support strategies for balance maintenance

Change in support strategies for balance recovery involves two different categories: stepping responses and reach-to-grasp strategies (Maki and McIlroy 1997).

Stepping responses are elicited when a postural perturbation is so strong that it causes the centre of mass to be displaced outside the base of support between the feet

and the ground (Luchies et al., 1994; Hsiao and Robinovitch, 1999; Wojcik et al., 1999), or simply by preference (McIlroy and Maki, 1993). Change in support strategies increase the base of support, and the range where the centre of mass can be displaced without a loss of stability (Maki and McIlroy, 1997)

Reach-to-grasp responses often involve a more precise sequence of kinematics, and require more complex control than stepping responses (Gage et al., 2007). One important distinction between reach-to-grasp strategies and stepping responses is that reaching parameters depend on the spatial representation of the object to be grasped relative to the body location (Desmurget et al., 1998), while during stepping responses, the target is usually constant and large (the ground) (Maki and McIlroy, 1997). To reach and grasp a handhold it is necessary to coordinate the upper limb in three dimensions at a speed of execution similar to that observed during stepping (Gage et al., 2007).

Reach-to-grasp responses involve three components: (1) transport of the hand to the target location, (2) flexing of the fingers to secure a hold on the support, and (3) grasp force development (Fan et al., 2005). Fan et al. (2005), examined the trajectory of the hand in various reach-to-grasp movements, and found that independent of the object location, the velocity of the limb was characterized by a bell-shaped curve, that is characteristic of voluntary movements (Corradini et al., 1992). They also found that the terminal velocity to reach the target did not change, independent of the target location.

However, there are differences in the dynamics governing volitional reach-to-grasp movements while standing on a stable surface and reach-to-grasp balance recovery responses following a sudden whole body perturbation (Maki, 2004). Maki and

McIlroy (1994) examined reach-to-grasp responses during whole body perturbation using a translating platform. Shoulder muscle activation occurred in 85% of the cases, following a delay of 90 to 140 ms similar to triggered responses of ankle muscles (Carpenter et al., 2005). They also found that arm responses were modulated accordingly to the perturbation direction and magnitude, even on the subjects' first trials. The authors concluded that muscle activation did not represent a generic startle response, but was instead driven and shaped by sensory feedback of the perturbation conditions.

Ghafouri and McIlroy (2004) tested the effect of vision on the reaching component of grasping during whole body perturbation. Different horizontal platform perturbations were applied to standing subjects, while vision was occluded during the initial response phase (within 200 ms following onset of the perturbation). The authors found that the early stage of hand transportation (100 ms) was unaffected by occlusion of vision, indicating that the central nervous system utilizes a combination of a pre-constructed visual spatial map (pre-visual map) and online feedback from somatosensory and/or vestibular mechanisms, to modulate the arm trajectory to the target.

1.4 The grasping hand

Napier (1956) was among the first to describe the biomechanics of grasping of the human hand. He described two general classifications for gripping an external object by the hand: the power grip and precision grip. The power grip involves activation of the

large wrist flexor muscles and allows the hand to oppose forces applied in any direction, while precision grip involves small adjustment of the fingers to control the direction of the forces applied to the object.

The posture of the thumb plays an important role in differentiating the power and precision grips. During the power grip, there is adduction of the metacarpo-phalangeal and carpo-metacarpal joints at the thumb (Napier, 1956). On the other hand, during the precision grip, these two joints are abducted and medially rotated at the carpo-metacarpal and the metacarpo-phalangeal joint. The changes in the thumb from a “wrapping” to an abducted position decreases net force generation, but increases precision in the location and direction of force application, facilitated by information from skin receptors (Napier, 1952). In the present study, I focus on the power grip, since this is most commonly used when standing on public vehicles and holding a handrail.

Pheasant and O’Neil (1975) proposed that the torque produced when holding an object depends on the coefficient of friction between the hand and the object, the net grip force (N), and the object dimensions. Seo et al. (2007) refined this model in proposing that torque (T) is given by the equation $T = \sum_{i=1}^n r_i \times \mu_i F_{ni}$, where n is the number of hand segments (such as the phalanges) in contact with the object, F_n is the normal force applied to the object by that hand segment, μ is the coefficient of friction between the object and the hand segment, and r is the distance from the object’s centre of mass to the point of force application by the hand segment. These authors found that torque generation depended on force direction (with higher values observed when frictional forces were directed towards the fingertips), and on object size. In particular, they found that lower torques were generated for larger handrail diameters. They

concluded that this is due not only to muscle length-strength relationships, but also to a decrease in the reaction force on the palm due the wider open aperture between the thumb and the fingers, and the generation of opposing forces in the fingers.

1.5 The effect of force direction and handrail height on hand force generation

Handrails in passenger vehicles vary in their design and configuration allowing for grasping at different heights (ranging from waist-height to overhead) and with different hand configurations to grasp vertical versus horizontally oriented handrails. Handrail configuration may influence the ability of passengers to resist perturbations in the anterior-posterior and medial-lateral directions, through the generation of forces at the hand. Indeed, several previous ergonomic studies have shown that handrail configuration influences the maximum force that can be generated during pushing or pulling (Jongkil, 2006; De Looze et al., 2000; Haslam, Boocock, Lemon, & Thorpe, 2002; Resnick & Chaffin, 1996; Snook, 1978; Snook & Ciriello, 1991).

However, these studies do not provide consensus on the most ergonomically favourable configuration (Hoozemans et al., 1998). For example, van der Beek et al. (2000) simulated work practices of postal workers and found that maximum forces for pushing were higher than pulling. On the other hand, Kumar (1995) found that pulling strength was higher than pushing strength. The difference may relate to experimental design: in Kumar's experiment, subjects were standing with the feet positioned consistently, and had to push and pull a handrail located in three different heights, while in Van der Beek's study, subjects were allowed to assume a free posture.

Age also alters grip strength. Uetake and Shimoda (2006) examined the ability of young and older adults to grip a cylindrical handrail (typical of the type found on buses) while sitting. Subjects sat and held a handrail and were subject to pulling and pushing forces of different magnitudes and directions. The authors found that maximum grip strength of elderly was on average 70% of younger adults. They also they showed that, independent of age and gender, strength was higher when pushing versus pulling the handrail. They also found that subjects could generate higher forces when holding 33 mm diameter than 42 mm diameter handrails. Unfortunately, these results are limited to the context of sitting.

Handrail height also affects subjects' ability to generate hand forces with subjects being able to generate less force when the handrail is located far from the being's centre of mass (Garg et al., 2005; Kumar, 1995; MacKinnon, 1998; Lee et al., 1991). For example, Kumar (1995) found that, during standing, males were able to generate on average 6% less and females 9% less hand force during pushing and pulling activities at high height (150 cm from the floor) compared to activities where the handrail was close to their centre of mass (100 cm). MacKinnon (1998) observed more dramatic differences with pulling forces being 47% lower when grasping a 200 cm high than a 107 cm high handrail.

1.6 Balance maintenance by grasping during whole body perturbation

Several previous studies have examined postural responses when the body is perturbed which the subject is grasping a handrail.

Noe and Martin (2003) asked subjects to rock on their heels with and without a hand support. The addition of handrail grasping changed the muscle activity patterns (EMG) in both the upper and lower limbs. Anticipatory postural adjustments were diminished in the lower limbs, showing reorganization in postural strategy between the two contexts.

Cordo and Nashner (1982) analyzed postural adjustments associated with whole body perturbations through a series of experiments where subjects stood on a platform that could move forward or backwards. In some trials, subjects were asked to hold a handrail in front of them while the platform moved, and in others subjects had to pull or push the handrail with the platform stationary. In trials with the platform moving, grasping again resulted in lower muscle activations in the lower limbs, showing a dependency of muscle activation on postural context. Also, during trials where subjects pulled the handrail with the stationary platform, lower limb activation preceded upper limb activation to maintain balance, illustrating a now well-studied anticipatory postural adjustment.

Jeka and Lackner (1994) showed that the simple act of touching a fixed experimental object can influence upright posture during stance. These authors tested subject's COP displacement during the Romberg stance in three conditions: not touching, fingertip touching and grasping a handrail mounted on the side of the subject. They found that both touch contact and grasping reduced COP displacement, compared to the non-touch condition. The authors concluded that fingertip contact provides information to the brain about body sway, and muscle activation is performed in either on anticipatory or reactive manner to reduce this body sway.

Slijper and Latash (2000) tested the effect of finger touch and hand grasp on anticipatory postural adjustments on standing subjects. Subjects were asked to flex the arm back and forth while standing on a stable or unstable surface. Finger touch decreased anticipatory postural adjustments (APA) in the leg and trunk muscles, but small changes were observed in the arm muscles. The authors concluded that APA's in leg muscles are more sensitive to perception (postural context) while arm muscles are more sensitive to mechanical aspects of posture.

1.7 Summary and goals

Public transportation continues to grow in relevance, vehicles such as buses, subways, and trains pose significant balance challenges for standing passengers, and injuries due to falls are common. The primary mechanism for fall prevention in standing passengers is handrail grasping. Yet, few ergonomic, biomechanical, or physiological studies have examined balance maintenance by handrail grasping in the context of improved handrail design for public transport vehicles, particularly in the area of fall and fall related injuries while riding in these vehicles. This thesis presents the results of laboratory experiments to improve our understanding of balance maintenance by handrail grasping, and the design of safer vehicle interiors.

The general goal of my thesis was to conduct laboratory experiments with young subjects to test the effect of two ergonomic factors: handrail configuration and stance configuration (examined via changes in perturbation direction), and one environmental

factor: perturbation magnitude, on the muscular effort involved in balance recovery during simulated vehicle starts and stops.

In particular, I conducted laboratory experiments to determine how the musculoskeletal demands of balance recovery were affected by:

- (a) The handrail grasp location (shoulder height versus overhead);
- (b) The stance configuration (anterior-posterior versus medial-lateral to the direction of platform translation);
- (c) The magnitude of the perturbation (low versus high).

Based on biomechanical considerations, I hypothesized that:

(1) Grasping an overhead versus shoulder height handrail reduces muscular effort. An overhead grasping configuration will result in lower magnitudes of centre of pressure excursion and hand contact force than observed for a shoulder height handrail (due to the larger moment arm of the hand contact force, with respect to the ankle).

(2) Adopting a sideways stance reduces muscular effort. Medial-lateral versus anterior-posterior stance configuration will result in lower (normalized) magnitudes of centre of pressure excursion and hand contact forces (due to the larger base of support provided by the feet).

(3) High perturbation magnitude will be more demanding than low perturbation magnitudes, resulting in higher centre of pressure excursions, hand contact forces and muscle activation.

Improved knowledge gained from this study will help to guide the design of safer vehicles, and enhance our understanding of the biomechanical demands involved in safety riding these vehicles.

CHAPTER 2: MATERIALS AND METHODS

2.1 Participants

Sixteen healthy young adults (8 men and 8 women) participated in this study (mean age = 26 yrs, SD = 5.3). All participants were recruited through advertisement and flyers at Simon Fraser University notice boards. Participant inclusion criteria included: (1) to be able to read and understand simple questions in English, (2) be older than 19 and younger than 35 years of age, (3) to be right side dominant, (4) have no known visual impairment, musculoskeletal condition or neurological disease.

The experiment protocol was approved by the Office of Research Ethics at Simon Fraser University and all the participants provided written informed consent.

2.2 Experimental protocol

The experiment involved three conditions (Fig. 1): handrail configuration, perturbation magnitude and perturbation direction. Both handrail configuration and perturbation magnitude had two levels (overhead and shoulder height grasping and high and low perturbation magnitudes respectively). Perturbation direction had four levels: forward, backward, sideways to the right and sideways to the left perturbations. During forward and backward (anterior-posterior) perturbations, the subject stood facing forward to the main plane of platform movement (Figs. 2A & 2C), while sideways

perturbations (lateral), the subjects were facing sideways to the main plane of movement. (Figs. 2B & 2D).

A linear motor (Trilogy, model T4DB80) of 1.2 m length was used to horizontally translate a customized platform secured on the top the motor, allowing us to mount a 32 mm diameter handrail (simulating those found in buses), in two configurations commonly observed in public transportation vehicles: overhead (at 1.75 m height), or at shoulder height (Fig. 2). Participants were asked to stand on the platform and hold the handrail lightly before the onset of the perturbation.

In each trial, the platform translated a total distance of 0.6 m. in the low magnitude perturbation, involving a peak acceleration of 1.0 m/s^2 , the peak velocity was 0.8 m/s. In the high magnitude (2.0 m/s^2) perturbations, the peak velocity was also 0.8m/s.

Prior to commencing a trial, subjects were informed that the platform would move randomly forward or backward (or left of right in the sideways stance configuration), with either high or low acceleration. They were also instructed to lightly hold the handrail prior to the perturbation (monitored via computer). After the perturbation onset, they could use as much hand force as necessary to prevent them from falling without taking a step.

Several studies examining postural response to platform perturbation have shown that a learning process occurs over the first few trials, and subjects are less stable during their first trials (Keshner et al., 1987; Marigold et al., 2002; Nijhuis et al., 2010). Since we were interested in simulating everyday responses with subjects responding naturally as riding a bus or subway, four practice trials were allowed for each hand vs.

stance configuration before the beginning of data collection (in a total of 16 trials). Every subject repeated each of the 16 experimental conditions three times, for a total of 48 trials.

In the present study, the term “forward perturbation” implied that the platform moved forward, inducing backward body sway. On the other hand, “backward perturbation” implied that the platform moved to the backward direction, inducing forward body sway. “Right perturbation” means that the participant was facing sideways to the main plane of movement and the platform moved to the right, inducing left body sway, while “left perturbation” means that the platform moved left, inducing a right body sway. The expected effects of perturbation direction on the mechanisms of balance recovery are shown in Fig. 3.

2.3 Data collection

In each trial, we measured the centre of pressure (COP) between the feet and the ground from a force plate (Bertec, Model 4060H) mounted flush on the top of the platform (Figures 1 and 3). We also measured hand contact force from a 3D load cell (Kistler, Model 9047C) secured to the base of the handrail. Finally, we measured muscle activity (Noraxon Myosystem 1200) from biceps brachii, triceps brachii, medial gastrocnemius and tibialis anterior muscle on the subjects’ right side. The platform acceleration was measured with an accelerometer (Kistler 8312A2 K-Beam) mounted on the top of the platform. This same accelerometer was used to calculate the platform

onset perturbation, defined as the instant that the acceleration exceeded 4 times the standard deviations above the baseline level in the preceding 500 ms interval.

All force signals as well as muscle activity were acquired at 2 KHz (see Figure 4, 9 and 10 for sample traces). Force signals were low-pass filtered recursively with a 50 Hz cut-off frequency, while muscle activation signals were full wave rectified and low-pass filtered (Butterworth 4th order with a 20 Hz cut-off frequency).

2.4 Data analysis

Outcome variables from each trial included: peak of COP excursion (PEAK_COP), peak of hand force generation (PEAK_FORCE), percentage of maximum COP displacement (COP%Max), percentage of maximum hand force (Force%Max), EMG integral (mv*s) of biceps brachii (BI_INT), triceps brachii (TR_INT), gastrocnemius (MG_INT) and tibialis anterior (TA_INT). To obtain the percentage of maximum COP displacement, prior to the data collection, each participant was asked to lean their body forward, backward, to the left and to the right as much as they could, without taking a step. Maximum values of COP displacement for each direction averaged over three repeated trials were considered 100%. The peak value they obtained (PEAK_COP) during the trials divided by this value, and multiplied by 100 was defined as COP%Max. To obtain the percentage of maximum hand force (Force%Max), a similar method was used. In each handrail configuration, participants were asked to push and pull the handrail to all four directions (forward, backward, sideways to the left and sideways to the right) for three repeated trials (static maximum forces). The average maximum

values obtained were considered 100% and used to calculate Force%Max. The muscles activation magnitudes were calculated by integrating the rectified EMG over the time period between the onset of the perturbation and the time where the platform reached maximum velocity (Fig. 4). EMG data collection and analysis were restricted to forward and backward perturbations only. Muscle onset activation was defined as the instant that the increase activity acceleration exceeded 4 times the standard deviations above the baseline level in the preceding 500 ms interval for each muscle.

2.5 Statistical analysis

A completely randomization design was logistically difficult to achieve due to the complexity of changes needed on the handrail configuration between the shoulder height and overhead. Therefore the experiment was conducted with a split-split plot design. This analysis is appropriated when there are restrictions on randomization of experimental conditions, as in the present experiment. In conducting the experiment, we first randomly selected the handrail configuration between overhead versus shoulder height. We then randomly selected the perturbation direction between anterior-posterior versus sideways.

General linear model analysis of variance was used to test the effect of grasping configuration, perturbation direction, and perturbation magnitude on raw and normalized values of COP and hand force. Similar analyses were undertaken for the intensities and onset latencies of muscle activations, comparing only the forward versus backward perturbation directions. In all analyses, the total alpha was set to 0.05.

In addition, Pearson product moment correlation analysis (with $p < 0.05$) was performed to quantify the strength of association between peak hand force development (PEAK_FORCE) and peak centre of pressure displacement (PEAK_COP), as well as the association between the percentage of maximum force development (Force%Max) and percentage of maximum centre of pressure displacement (COP%Max).

CHAPTER 3: RESULTS

All subjects were able to perform the tests successfully and no steps or falls were observed. Figure 4 represents traces of the platform kinematics, COP excursion, hand contact forces and EMG of upper and lower limbs over 2.5 seconds for a single subject grasping the handrail at shoulder height during a forward perturbation (black lines) and a backward perturbation (gray lines). The first dashed line represents the perturbation onset ($t = 0$) where the platform starts to accelerate. The total platform distance traveled was 0.6 m from the initial position, the peak velocity was 1.2 m/s, and the peak acceleration was 2m/s^2 . Note that the second dashed line is located when the platform reaches the peak velocity (around 700 ms after the onset perturbation). At this point the acceleration is equal to “0” and the platform starts to decelerate (the acceleration starts to become negative). In the present study, we analyzed only the results from the first acceleration phase (values observed between both dashed lines).

3.1 COP excursion

Table 1 shows the values for mean and standard deviation for centre of pressure displacement (raw and normalized) for all the 16 different conditions. We found that peak centre of pressure excursion (PEAK_COP) was affected by grasping configuration, (Fig. 5 and Table 1) with PEAK_COP being 8% higher for overhead grasping compared to shoulder height grasping (mean values 9.2 cm versus 8.5 cm, respectively; $F = 5.6$, $p = 0.02$). PEAK_COP was also affected by perturbation magnitude, being 24% larger for

high magnitude than low magnitude perturbations (9.8 cm versus 7.9 cm, respectively; $F = 107.6$, $p < 0.001$) (Fig. 5-A3). PEAK_COP was also affected by perturbation direction ($F = 178.8$, $p < 0.001$), where on average, forward perturbations (which induced movement of the COP posterior to the ankle) involved approximately 59% lower values of PEAK_COP excursion when compared to backward perturbations (3.6 cm and 8.8 cm respectively), and approximately 78% lower values than right or left perturbations (Fig. 5-A2). Furthermore, there was a significant interaction between perturbation direction and perturbation magnitude ($F = 10.5$, $p < 0.0001$), with perturbation magnitude having a more striking effect on PEAK_COP in left and right perturbations, than in forward or backward perturbations (Table 1).

When the centre of pressure values were normalized (COP%Max), differences were found for perturbation direction ($F = 5.3$, $p < 0.01$) and magnitude ($F = 47.9$, $p < 0.001$), but not for grasping configuration ($F = 3.4$, $p = 0.07$) (Fig. 5, right column). COP%Max was 27% higher during backward perturbations than forward perturbation (83.2% versus 65.5%, respectively), reflecting a tendency to move more towards maximum forward than backward extremes of the COP. No differences were found between left and right perturbations. High perturbation magnitudes displaced the COP%Max 21% farther than low perturbation magnitudes (66.3% versus 80.3% respectively). Furthermore, there was a significant interaction between stance configuration and perturbation magnitude ($F = 2.9$, $p = 0.037$) (Fig. 6), with perturbation magnitude having a stronger effect on COP%Max in backward than forward perturbations.

3.2 Hand force

Table 1 shows the values for mean and standard deviation for hand forces (raw and normalized) for all the 16 different conditions. Peak of hand force development (PEAK_FORCE) was affected significantly by grasping configuration ($F = 29.2$, $p < 0.0001$), being 30% larger for shoulder-height than overhead grasping (58.2 N versus 44.6 N). PEAK_FORCE was 142% larger for higher compared to low perturbation magnitudes (72.8 N versus 30.0 N, respectively; $F = 236.7$, $p < 0.0001$). PEAK_FORCE also depended on perturbation direction ($F = 48.2$, $p < 0.0001$), averaging 49% smaller in sideways than forward, and 30% smaller in sideways than backward perturbation (Fig. 7-A2). Significant interaction was found between perturbation direction and grasping configuration ($F = 6.7$, $p < 0.001$), with forward perturbation during shoulder height grasping being on average 89% higher than all the other perturbation direction and grasping configuration (Fig. 8).

The percentage of maximum hand force development (Force%Max) was significantly different with grasping configuration ($F = 11.5$, $p < 0.002$), perturbation direction ($F = 5.1$, $p = 0.003$) and perturbation magnitude ($F = 138.4$, $p < 0.001$) (Fig. 7). Shoulder height grasping on average, required 33% more force from the upper limbs than overhead grasping (71.4% versus 54% of Force%Max, respectively). Forward perturbation required 34% and 55% more hand force compared to right and left perturbations (77.4% Force%Max for forward, 57.6% Force%Max for right and 50.0% Force%Max for left perturbation). No differences were found between left and right Force%Max neither forward versus backward Force%Max. The interaction analysis indicated that perturbation direction depended on grasping configuration ($F = 4.3$, $p <$

0.01), where forward perturbations while shoulder grasping demanded more hand force from the individuals (72%) compared with the other stance versus grasping configuration (Fig. 8). Also, high perturbation magnitudes required 157% more hand force compared to low perturbation magnitudes (90.1 versus 35.0% respectively).

The maximum force generated during the static trials was significant dependent on perturbation direction ($F = 4.7$; $p < 0.01$). More specifically, pulling the handrail generated 35 % higher forces than forces generated sideways (116 N for forward, 86 N for left and 85 N for right directions). No differences were found between overhead and shoulder height static force generation ($p = 0.14$).

The correlation analysis showed a significant inverse correlation between PEAK_FORCE and PEAK_COP ($r = -0.3$, $p < 0.0001$). However, when the data were normalized, no correlation was found between Force%Max and COP%Max ($r = -0.1238$, $p = 0.075$).

3.3 Muscle responses

Muscle activation patterns provide insight on the physiological mechanisms underlying changes in hand force and COP development. I focused on comparing responses in anterior versus posterior platform translations. Figures 9 and 10 show traces of kinematics, hand forces and EMG from typical trials for a single subject, while grasping a handrail at shoulder height and overhead, respectively. Following a forward perturbation (Figures 9A), which induces a backward fall, a short burst of increased activity occurred in the tibialis anterior, starting approximately 124 ms following the onset

of platform acceleration. This caused a posterior movement of the COP. Near simultaneously, there was commencement of a longer sustained burst of the biceps, with corresponding increases in hand force in an anterior (stabilizing) direction. In general, forward perturbation (Figures 9A and 10A) involved lower amplitudes of triceps and gastrocnemius activation, while backward perturbation (Figures 9AB and 10B) involved lower magnitudes of biceps and tibialis anterior activation, illustrating a selective of the motor synergy depending on perturbation direction. Indeed, Fig. 9B exhibits nearly complete silencing of the biceps and suppression of the early stage response in the tibialis anterior. Overhead grasping involved more biceps activation in backward falls.

Statistical analysis showed that BI_INT was significantly different between grasping configurations, being 51.3% higher for overhead grasping compared to shoulder height grasping (139.1 versus 91.9 mv*s, respectively; $F = 11.1$, $p < 0.005$) (Fig. 11). BI_INT was also dependent on perturbation magnitude, being 58.1% higher during higher magnitudes than lower magnitudes (141.5 versus 89.5 mv*s, respectively; $F = 42.1$, $p < 0.0001$). Finally, BI_INT depended on perturbation direction, being 157% higher for forward perturbations than backward (166.3 versus 64.7mv*s, respectively; $F = 51.4$, $p < 0.0001$). Similarly, TRI_INT was affected by grasping configuration ($F = 7.9$, $p = 0.007$), perturbation magnitude ($F = 33.3$, $p < 0.0001$) and perturbation direction ($F = 19.7$, $p < 0.0001$). TRI_INT was 39.2% higher for overhead than shoulder height grasping (54.7 versus 39.3 mv*s, respectively), 52.3% higher during higher than lower perturbation magnitudes (56.8 versus 37.3 mv*s, respectively), and 69.6% higher for backward than forward perturbation (59.2 versus 34.9 mv*s, respectively). TA_INT was also influenced

by grasping configuration ($F = 7.7$, $p < 0.01$) and perturbation direction ($F = 9.2$, $p < 0.005$), averaging 82.1% higher for overhead than shoulder height grasping (46.8 versus 25.7 mv^*s , respectively), and 93.5% higher during forward than backward perturbation (47.8 versus 27.7 mv^*s , respectively). Analysis of MG_INT showed differences only for perturbation direction, where muscle activity during backward perturbation was 75.5% higher than during forward perturbation (63.7 versus 36.3 mv^*s , respectively; $F = 15.6$, $p < 0.005$) (Table 1).

There was a strong effect of perturbation magnitude on muscle activation onset. BI_ON activation was 26% faster during higher than lower perturbation magnitude (190ms vs. 240ms, respectively), TR_ON activation was 41% faster (170ms vs. 240ms, respectively), TA_ON activation was 49% faster (185ms vs. 276 ms, respectively) and MG_ON activation was 20% faster (204ms vs. 244ms, respectively) during higher magnitudes (Table 1).

There was also an effect of perturbation direction on muscle activation onset. During forward perturbations BI_ON activation was 57% higher than backward activation (164ms vs. 263ms respectively, $F = 18.5$, $p < 0.001$), and TA_ON was 22% (208ms vs. 253 ms, respectively, $p < 0.04$; $F = 4.5$). On the other hand, during backward perturbations TR_ON activation was 18% faster (188ms vs. 222ms, respectively, $p < 0.04$; $F = 5$) and MG_ON activation was 45% faster (175ms vs. 253ms, respectively, $p < 0.001$; $F = 18.5$) than forward perturbations.

CHAPTER 4: DISCUSSION

Maintaining balance while standing on a bus, subway or train is a challenging task, and standing passengers often rely on handrail grasping to prevent falls, especially during vehicle starts and stops. This study examined the effect of handrail and stance configuration, and perturbation magnitude on two measures of the effort involved in balance maintenance: the peak force applied to the hand, and the peak excursion of the centre of pressure between the feet and the ground.

I hypothesized that, when compared to a shoulder height handrail, an overhead grasp configuration would result in lower magnitudes of centre of pressure excursion and hand contact force. My results partially support this hypothesis. I found that normalized values of hand force development (Force%Max) were indeed lower for the overhead than shoulder-height grasping configuration, but that values of normalized COP excursion (COP%Max) were unaffected by handrail configuration. I also found that muscle activities of biceps brachii, triceps brachii and tibialis anterior were generally higher (despite the fact that forces were lower) for overhead grasping than shoulder height grasping. This may relate to differences between postures in the roles of the biceps and triceps brachii in stabilizing the shoulder.

I also hypothesized that that, when compared to forward perturbations, sideway perturbations would involve lower magnitudes of centre of pressure excursion and hand contact forces. Again, my results provide only partial support for this hypothesis. I found that sideways perturbations indeed generated lower Force%Max compared to forward,

but no difference were found for COP%Max between these conditions. Additionally, when comparing forward versus backward perturbation, muscle activation intensities were selective with perturbation direction, being higher for BI_INT and TA_INT in forward perturbations, and higher in TR_INT and MG_INT for backward perturbations. Furthermore, there was quicker onset of activation (of the biceps) in forward perturbations, perhaps due to the smaller base of support posterior to the ankle, and need for faster upper limb force generation to help restore balance.

Finally, I hypothesized that an increase in perturbation magnitude would result in larger centre of pressure excursions, hand contact forces and muscle activations. I indeed found that COP%Max and Force%Max increased with perturbation magnitude, as did activation intensities for the biceps, triceps and tibialis anterior (but not the gastrocnemius). Furthermore, higher perturbation magnitudes caused faster onsets of muscle activation, for all lower and upper limb muscles.

My results agree with others on the effect of perturbation direction on hand force development during grasping. Similar to Uetake and Shimoda (2006), I found that subjects were able to generate higher static forces while pulling the handle backwards than while pushing sideways. Also, peak observed forces were higher when pulling (during forward perturbations) than in sideways perturbations, in both their study and mine.

My results also complement previous studies examining how handrail location influences force development. Both Garg et al. (2005) and Kumar (1995) reported that maximum attainable hand forces in a standing grasping task were lower for overhead

than shoulder height handrails. However, for the handrails I examined, there was no effect of handrail height on maximum force generation, or on peak observed forces.

I found that the sequence and magnitude of muscle activations for balance recovery was selective to the perturbation direction and magnitude. Forward perturbations (backward falls) involved greater activation and faster onset of biceps brachii and tibialis anterior. Conversely, backward perturbations (forward falls) involved greater activation as well as faster onset of triceps and gastrocnemius. I also found that onset latencies were also 26-44% faster in high than low magnitude perturbations trials. These observations agree with Horak and Nashner (1986), who showed that the sequence of muscle activations during balance recovery by grasping depends on the direction and magnitude of the perturbation. Similar results were reported by McIlroy and Maki (1995) for the transport phase of reach-to-grasp responses.

I also found that stance configuration influenced the muscular demands on the upper versus lower extremities. During backward falls (caused by forward perturbations), participants relied primarily on an upper limb strategy, involving higher hand forces and lower COP displacements. Conversely, sideways perturbations elicited more of a lower limb strategy. This observation underlines the need for a detailed examination of the characteristics of real-life perturbations, to assist in interpreting the current results.

These results support previous studies in providing evidence of a relative weighting between upper and lower limb control depending on perturbation direction, magnitude, and use of handrails for support. This weighting is likely mediated by propriospinal pathways between the lumbar and cervical enlargements of the spinal cord

(Marigold and Misiaszek 2009).

No previous studies, to my knowledge, have measured both hand forces and muscle activities during balance recovery by grasping. I found that each provide complementary information. For example, I observed that forces were lower but activation intensities were higher for overhead than shoulder height handrails. I also found that raw versus normalized values of hand force and COP excursion provide relevant information. Normalized values of COP excursion were not different between the four perturbation directions, but raw magnitudes were higher for sideways and backward than for forward. This later observation agrees with Henry et al (1998), and reflects the fact that the attainable COP excursion is much smaller posterior to the ankle than anterior or in the medial-lateral direction.

My study had important limitations. I examined simulated vehicle starts and stops in a laboratory environment using a linear motor having a travel of 1.2 m. I incorporated peak accelerations typical of those measured previously on buses and subways. Thus, my results may not reflect strategies used in slower stop and starts, or perturbations applied over a larger distance. I also analyzed only two handrail locations (shoulder height and overhead) and a single stance width (based on shoulder width) while in real life, passengers may grasp handrails over a wide range of heights and adopt a wide range of stance configurations. I also tested only one handrail diameter with specific frictional characteristics. Alterations in these variables will influence passenger's ability to generate hand forces. I analyzed only vehicle starts and stops, and not turns. I also normalized the peak COP excursion to "perceived" limits of stability (measured in maximum lean tests) and some of the observed variability may relate to differences

between subjects (or between conditions) in perceived base of support, instead of actual base of support. Similar considerations may exist for maximum perceived hand force. Also, I did not examine passenger fatigue issues due to long-term grasping (which one would expect to be more stressful in overhead than shoulder-height grasping).

In summary, hand forces were highest when grasping shoulder-height handrails, and in forward perturbations. Conversely, COP excursion was highest in sideways and backward perturbations. This reflects that participants relied more on COP excursion than hand force to maintain balance during backward and sideways perturbations, while the opposite was true for forward perturbations. Practically, a forward stance configuration will result in perturbations in both forward and backward directions (during starts and stops, respectively), while a sideways configuration largely eliminates this directional sensitivity. Collectively, these observations indicate that a combination of overhead handrail grasping and sideways stance is most effective in minimizing the muscular effort associated with balance maintenance while standing on public transportation vehicles. However, additional studies are required to test this hypothesis in the real-life environment.

CHAPTER 5 THESIS SYNTHESIS AND FUTURE DIRECTIONS

The primary goal of this thesis research was to examine how the muscular effort involved in restoring balance while standing and grasping (as is common while riding public vehicles) depends on stance configuration, handrail configuration, perturbation direction, and perturbation magnitude. This information is important for the design of safer and more comfortable buses, subways and trains.

Collectively, my results indicate that a combination of overhead handrail grasping and sideways stance is most effective in minimizing the muscular effort associated with balance maintenance while standing on public transportation vehicles. This information may contribute to the design of safer vehicles interiors (handrail location), improved passenger information on how to safely ride vehicles (stance configuration) and improved guidelines on driver behaviour (acceptable vehicle accelerations thresholds).

However, many important questions remain unanswered, and motivate future studies. These include an examination of muscle fatigue and comfort when holding a handrail with different postures, over long durations. Studies are also warranted on the effect of using the dominant versus non-dominant hand (or both hands) for grasping. An examination also is warranted on balance maintenance through grasping overhead straps (a common vehicle feature).

Another important topic for future studies is the effect of age on optimal handrail and stance configurations, and ability to maintain balance by handrail grasping. Several questions are of interest. For example, how are COP excursions and hand forces

affected by aging? Are different strategies (i.e., roles of upper versus lower limb responses) used by older versus younger populations to maintain balance by handrail grasping? What is the relative importance of declines in strength versus reaction time in accounting for such differences (Robinovitch et al., 2002; Mackey and Robinovitch, 2006)?

Also of interest is the effect of different handrail materials on ability to restore balance. Normally, the handrails inside public transportation vehicles have a smooth aluminum surface. A higher friction material may enhance torque generation and the stabilizing effect of hand force generation. Furthermore, a foam layer may reduce risk for injury if the passenger impacts a handrail, but reduce sensory information from cutaneous mechanoreceptors in the fingers important to grasping responses. Further experiments are required to explore these issues.

Finally, another important direction for future research is measurement of balance maintenance strategies in real life (moving vehicle) situations, perhaps in collaboration with vehicle manufacturers and/or local transit authorities. This requires the use of miniature wearable sensors (accelerometers, gyros, and perhaps EMG) to measure postural responses, and handrails instrumented with load cells to record hand forces. While less controlled, the external validity of these measures makes them an essential complement to laboratory experiments.

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Tables

Table 1: Means and standard deviations (SD) of centre of pressure (COP) and hand forces for the various hand configurations, perturbation directions and perturbation magnitudes.

Hand configuration	Overhead								Shoulder							
	Low				High				Low				High			
	Perturbation direction	For	Back	Left	Right	For	Back	Left	Right	For	Back	Left	Right	For	Back	Left
Peak COP displacement (cm)	3.8 (1.7)	8.3 (1.5)	11.0 (1.9)	9.2 (2.0)	4.3 (1.7)	10.3 (1.9)	13.9 (3.3)	12.8 (1.8)	3.1 (1.8)	7.8 (2.4)	10.6 (1.7)	9.6 (2.2)	3.2 (2.1)	9.0 (2.1)	13.2 (2.1)	11.8 (2.2)
Peak hand force (N)	40.6 (14.5)	25.6 (10.7)	15.5 (10.0)	17.5 (8.2)	82.0 (25.8)	68.7 (28.8)	57.4 (34.6)	49.4 (18.7)	56.5 (23.5)	41.9 (18.7)	14.1 (9.4)	28.6 (23.8)	118.7 (48.8)	82.9 (38.3)	49.7 (22.6)	73.6 (33.3)
COP% max (%)	68.0 (27.0)	78.7 (11.4)	69.1 (13.3)	55.8 (14.8)	77.1 (30.6)	94.6 (14.5)	86.4 (21.2)	80.2 (19.6)	58.2 (40.1)	72.3 (21.4)	66.8 (14.5)	59.6 (14.8)	56.5 (31.6)	85.4 (20.3)	83.9 (22.8)	74.4 (19.4)
Hand force% max (%)	39.1 (17.4)	30.5 (20.7)	21.0 (18.0)	28.5 (17.9)	75.9 (43.3)	78.3 (51.7)	89.7 (75.3)	73.9 (44.7)	61.1 (41.1)	45.0 (24.5)	19.5 (13.3)	39.2 (39.3)	138.1(8 0.9)	109.8(4 9.5)	73.6 (47.2)	90.3 (55.8)
BI_INT (mv*s)	158 (78)	51 (56)	-	-	247 (121)	101 (106)	-	-	103 (54)	46 (49)	-	-	157 (76)	62 (57)	-	-
TR_INT (mv*s)	33 (22)	57 (44)	-	-	54 (43)	75 (35)	-	-	22 (12)	37 (19)	-	-	31 (16)	68 (36)	-	-
TA_INT (mv*s)	58 (50)	37 (54)	-	-	64 (59)	28 (33)	-	-	31 (34)	18 (23)	-	-	38 (58)	17 (19)	-	-
MG_INT (mv*s)	28 (22)	67 (28)	-	-	54 (53)	70 (45)	-	-	26 (16)	65 (46)	-	-	37 (19)	53 (39)	-	-
BI_ON (ms)	219 (105)	273 (188)	-	-	137 (32)	277 (143)	-	-	174 (85)	295 (202)	-	-	135 (39)	207 (182)	-	-
TR_ON (ms)	226 (102)	235 (95)	-	-	187 (70)	168 (55)	-	-	279 (136)	221 (97)	-	-	191 (76)	134 (54)	-	-
TA_ON (ms)	253 (116)	295 (168)	-	-	168 (42)	204 (89)	-	-	224 (175)	299 (185)	-	-	163 (63)	207 (88)	-	-
MG_ON (ms)	273 (215)	211 (63)	-	-	235 (108)	144 (40)	-	-	308 (179)	191 (44)	-	-	187 (86)	155 (41)	-	-

NOTES: For = forward; Back = backward; BI_INT = biceps brachii integrated amplitude; TR_INT = triceps brachii integrated amplitude; TA_INT = tibialis anterior integrated amplitude; MG_INT = gastrocnemius integrated amplitude; BI_ON = biceps brachii activation onset; TR_ON = triceps brachii activation onset; TA_ON: tibialis anterior activation onset; MG_ON: gastrocnemius activation onset

Figures

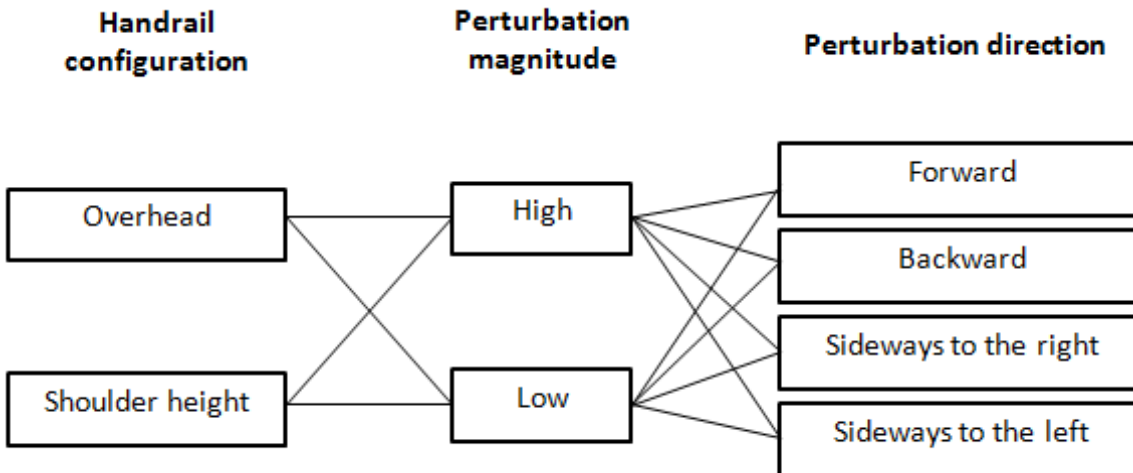


Figure 1: Schematic representation of the different experimental conditions. Forward and backward perturbations were evoked with the subjects standing upon the platform with their sagittal plane aligned with the platform X axis of movement. Sideways perturbations (to the left and to the right) were evoked with the participants standing upon the platform with their sagittal plane perpendicular to the platform X axis of movement.

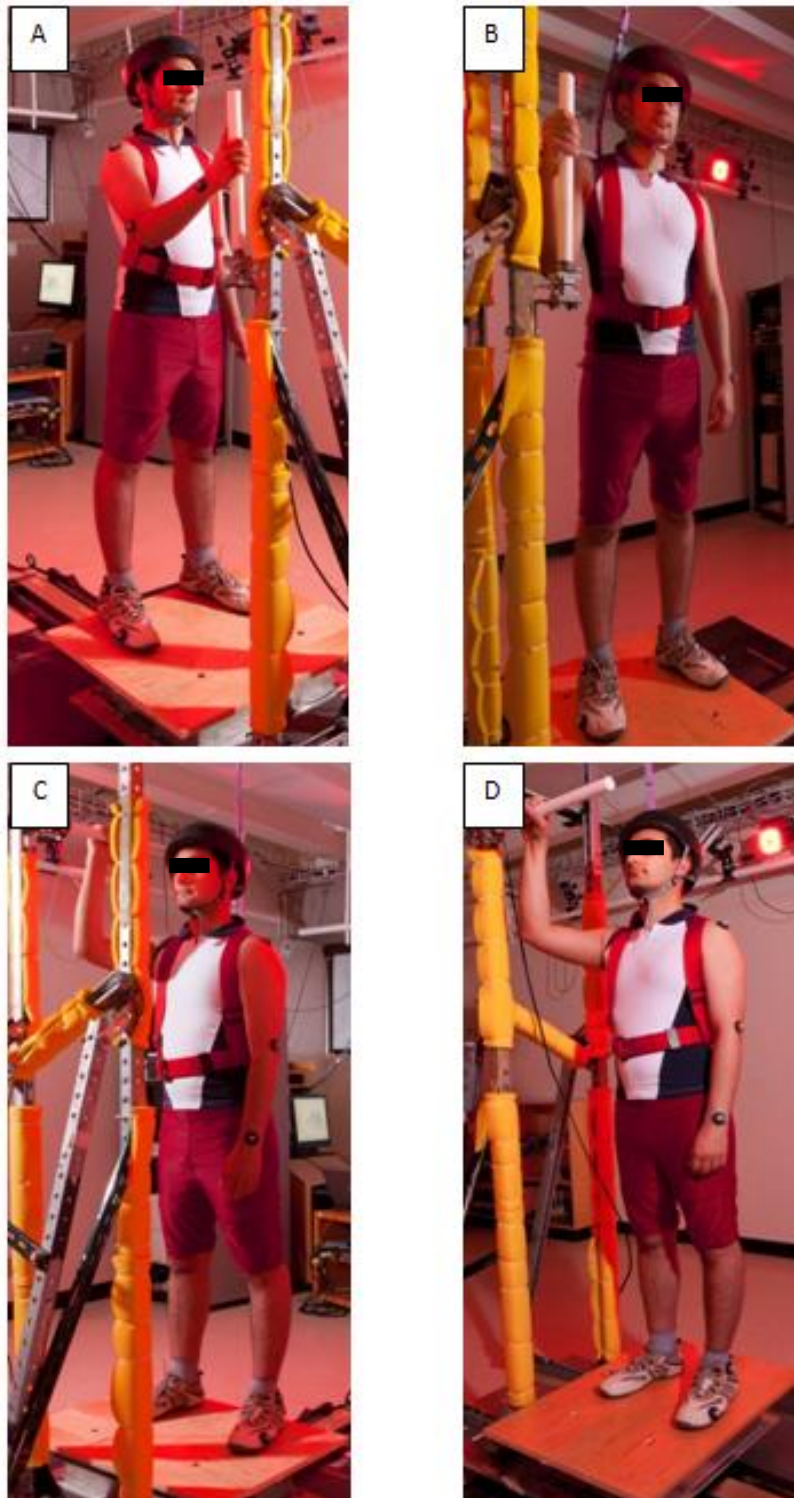


Figure 2: Experimental conditions. A customized structure secured to the linear motor simulates two grasping configurations: shoulder height or overhead. Perturbations were either forward/right or backward /left (depending on stance configuration). (A) Shoulder height grasping during forward/backward perturbations; (B) Shoulder height grasping during right/left perturbations; (C) Overhead grasping during forward/backward perturbations and (D) overhead grasping during right/left perturbations.

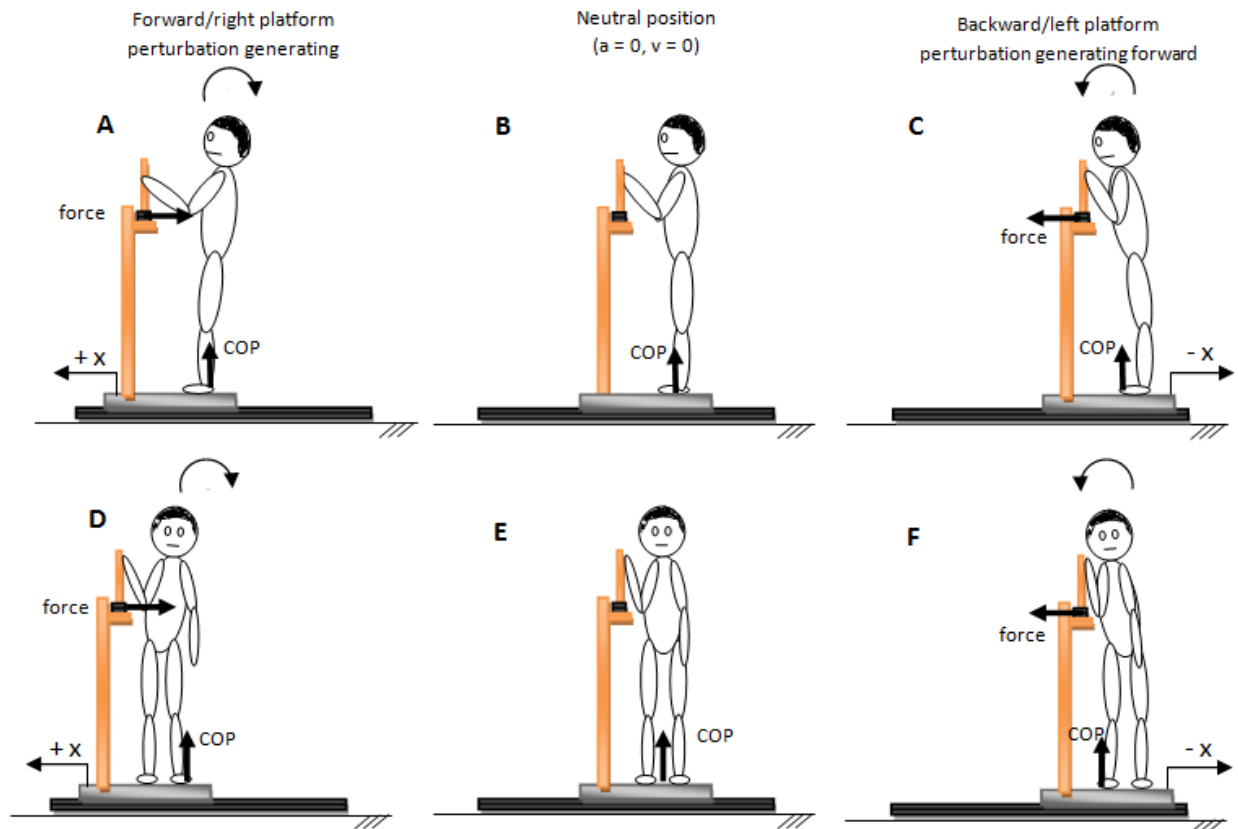


Figure 3: Effect of perturbation direction on recovery mechanisms of balance. (A) Forward platform perturbation generates backward body sway; (B) Neutral position for anterior-posterior perturbation; (C) Backward platform perturbation generates forward body sway; (D), right platform perturbation generates left body sway; (E) Neutral position for lateral perturbation; (F) Left platform perturbation generates right body sway. These perturbations are countered by the development of appropriate hand forces and COP excursions.

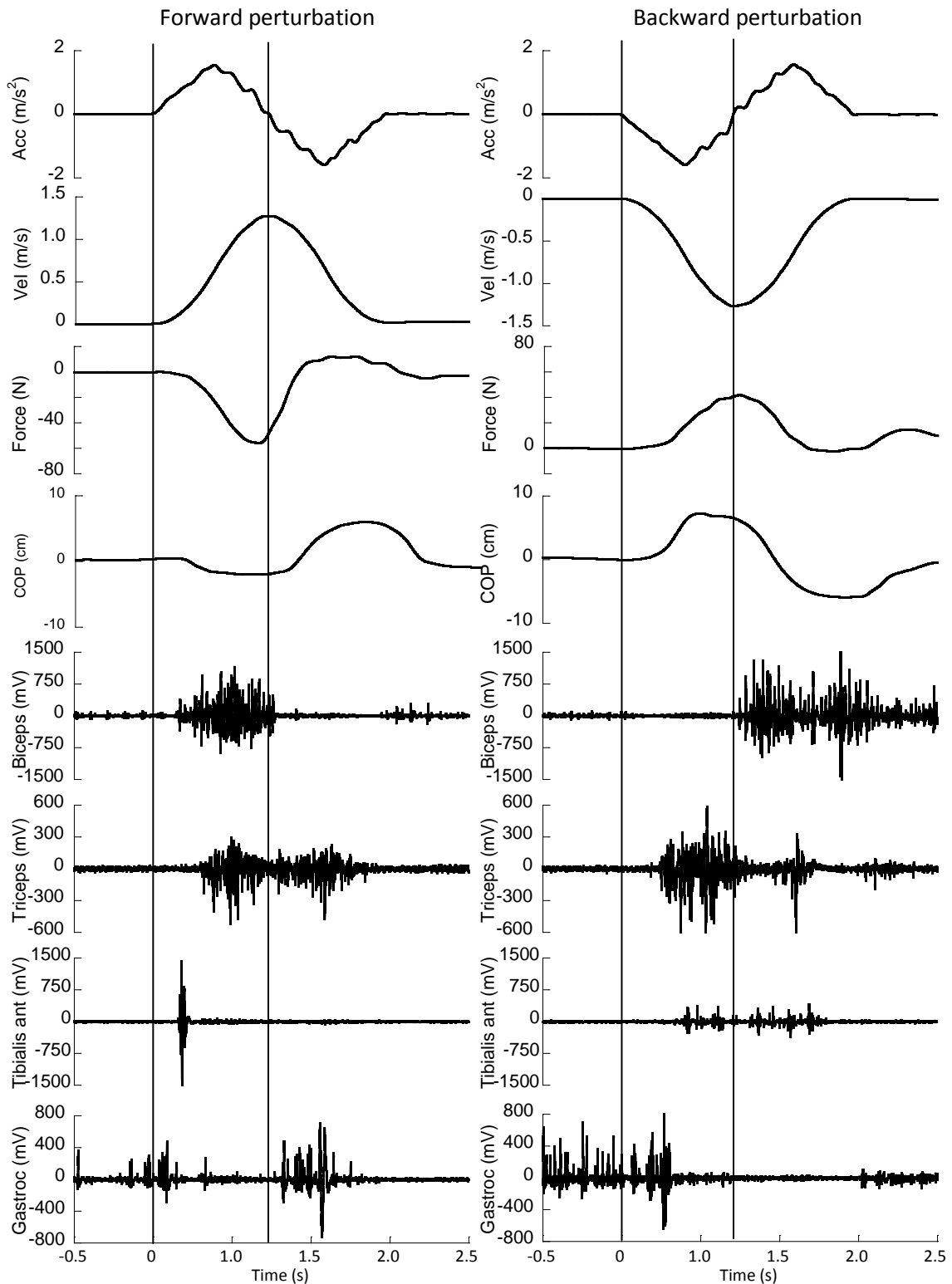


Figure 4: Traces of platform displacement, velocity, and acceleration, hand force, COP excursion and EMG activity over time, for a single subject holding the handrail at shoulder height, subjected to high magnitude perturbation (peak acceleration = 2 m/s²). Vertical lines show the onset of platform movement, and the time of peak forward velocity (and zero acceleration). A forward perturbation induces a backward fall, while a backward perturbation induces a forward fall.

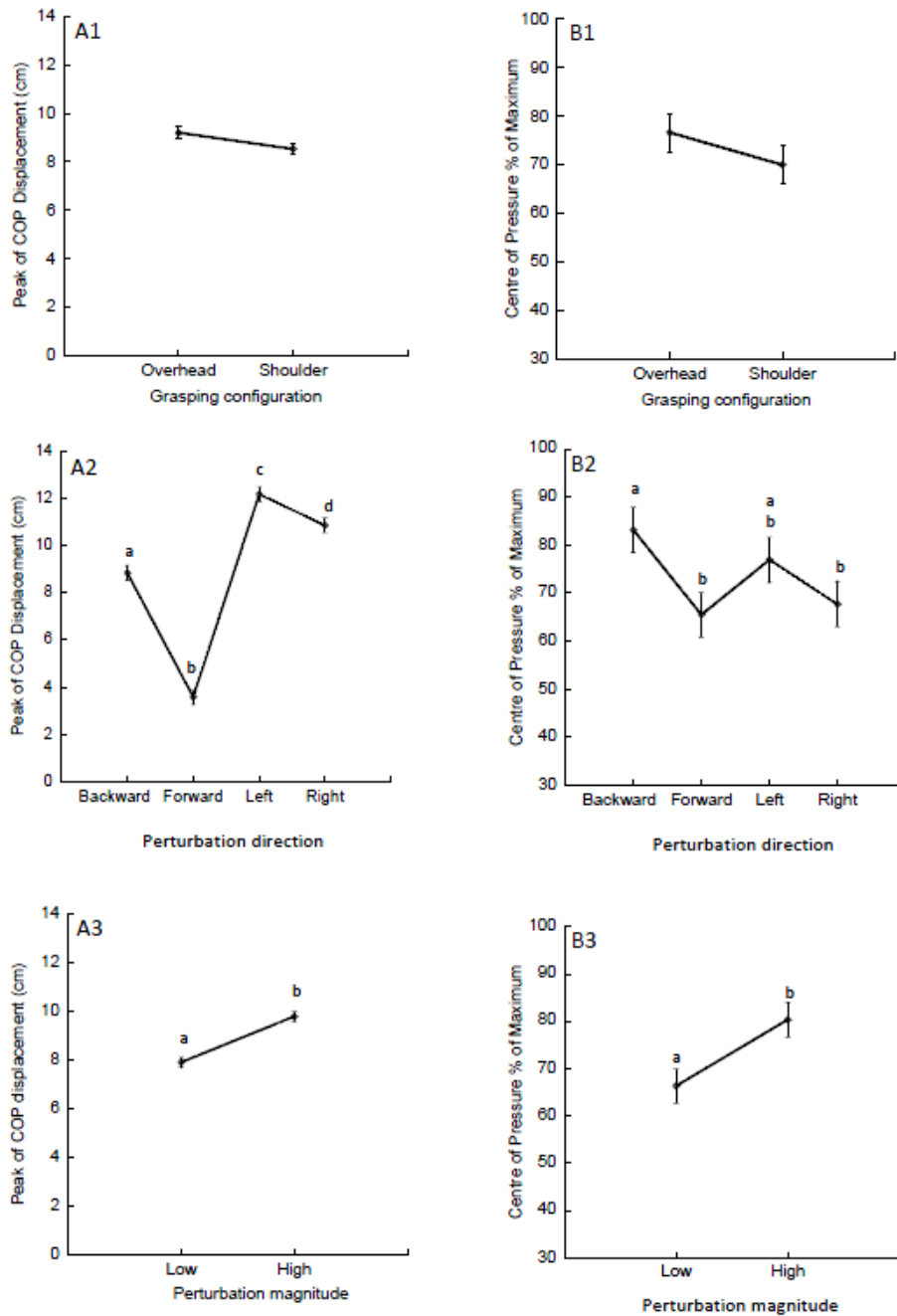


Figure 5: Peak centre of pressure displacement (raw and adjusted) for different perturbation directions and magnitudes. The first row shows the effect of: (A1) grasping configuration on peak COP displacement and (B1) grasping configuration on centre of pressure% of maximum. The second row shows the effect of: (A2) perturbation direction on peak COP displacement and (B2) the effect of perturbation direction on centre of pressure% of maximum. The third row shows the effect of: (A3) perturbation magnitude on peak COP displacement and (B3) centre of pressure% of maximum. Average values and standard errors (vertical bars) are shown. Levels not connected by same letter (i.e. *a* and *b*) are significantly different ($p < 0.05$). Analysis of raw data (left column) shows that higher peak of COP displacement occurred during sideways perturbation, however, during normalized data (right column) backward perturbation required more from the lower limbs compared to the other perturbation directions. It was also observed that higher the magnitude, higher the COP displacement for both raw (A3) and normalized data (B3).

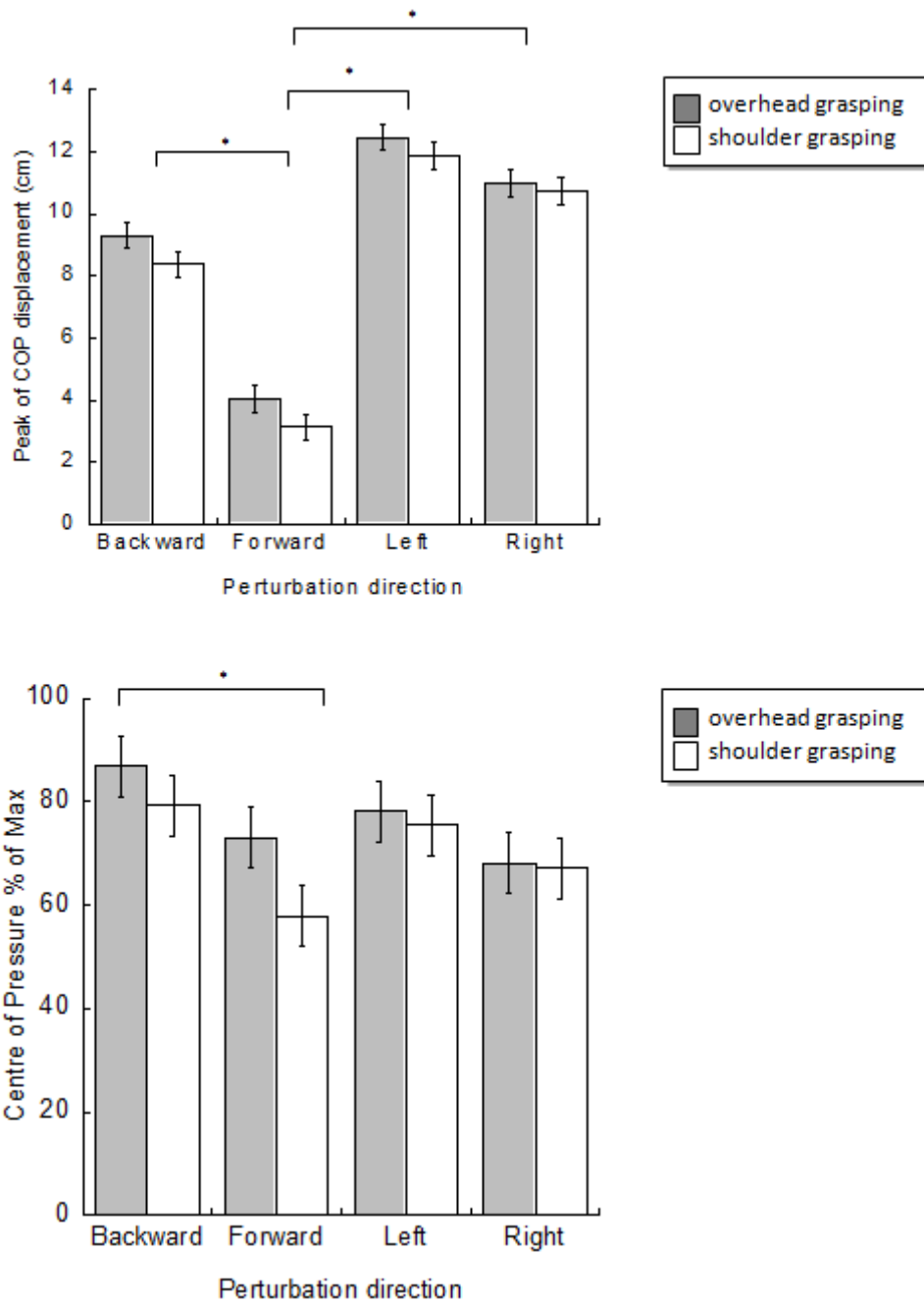


Figure 6: Effect of perturbation direction and hand configuration on (A) raw magnitudes of peak COP displacement and (B) peak of normalized hand force. Average values and standard errors are shown. Asterisks show where these differences occurred ($p < 0.05$).

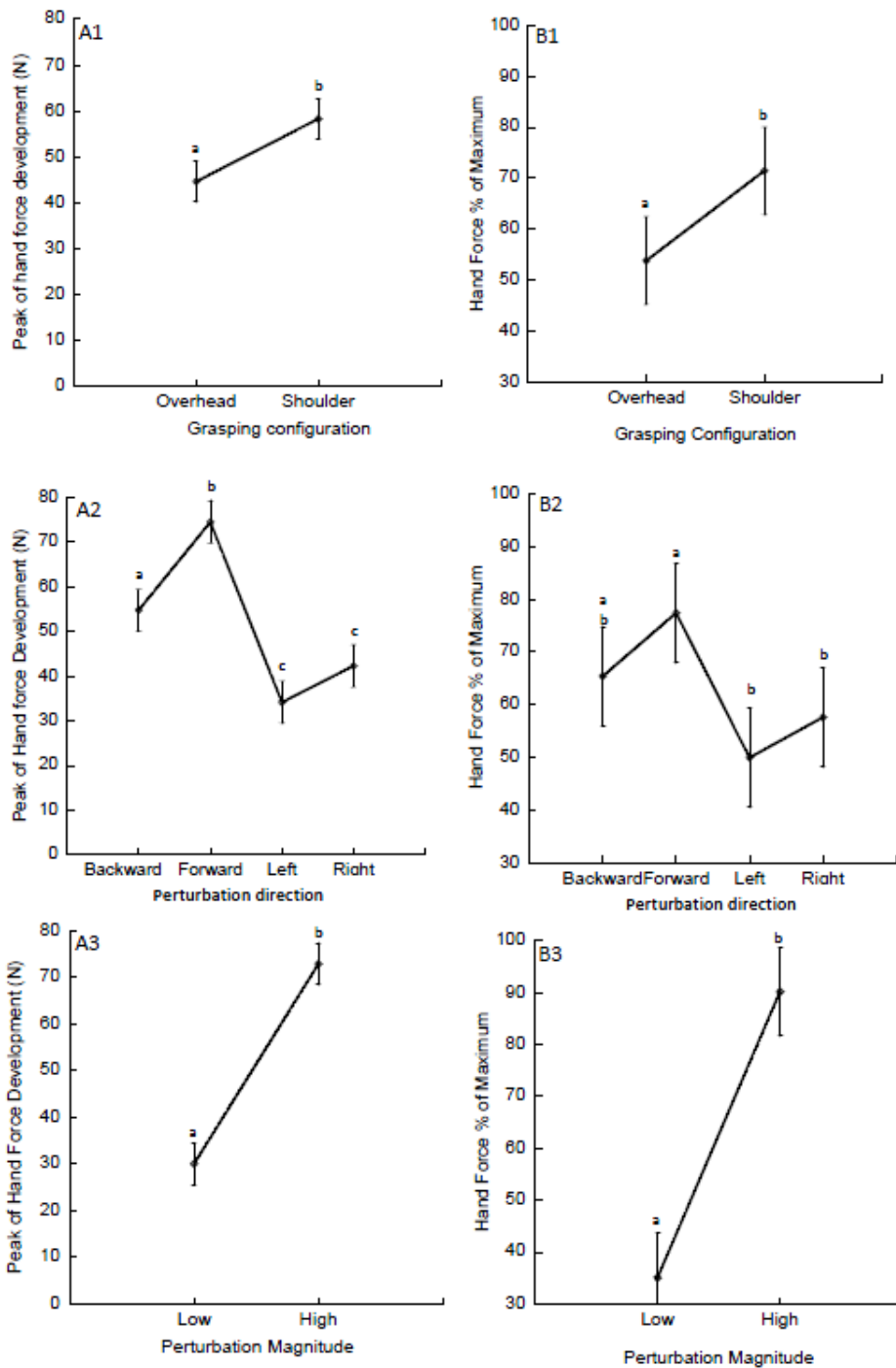


Figure 7: Raw and adjusted (left and right columns respectively) peak hand force for different grasping configurations, perturbation directions and perturbation magnitudes. The first row shows the effect of perturbation direction on hand forces (A1 raw, B1 adjusted). The second row shows the effect of perturbation magnitude on hand forces (A2 raw, B2 adjusted). The third row the effect of perturbation magnitude on hand force generation (A3 raw, B3 adjusted). Average values and standard errors (vertical bars) are shown. Average values and standard errors (vertical bars) are shown. Levels not connected by same letter (i.e. a, b and c) are significantly different ($p < 0.05$).

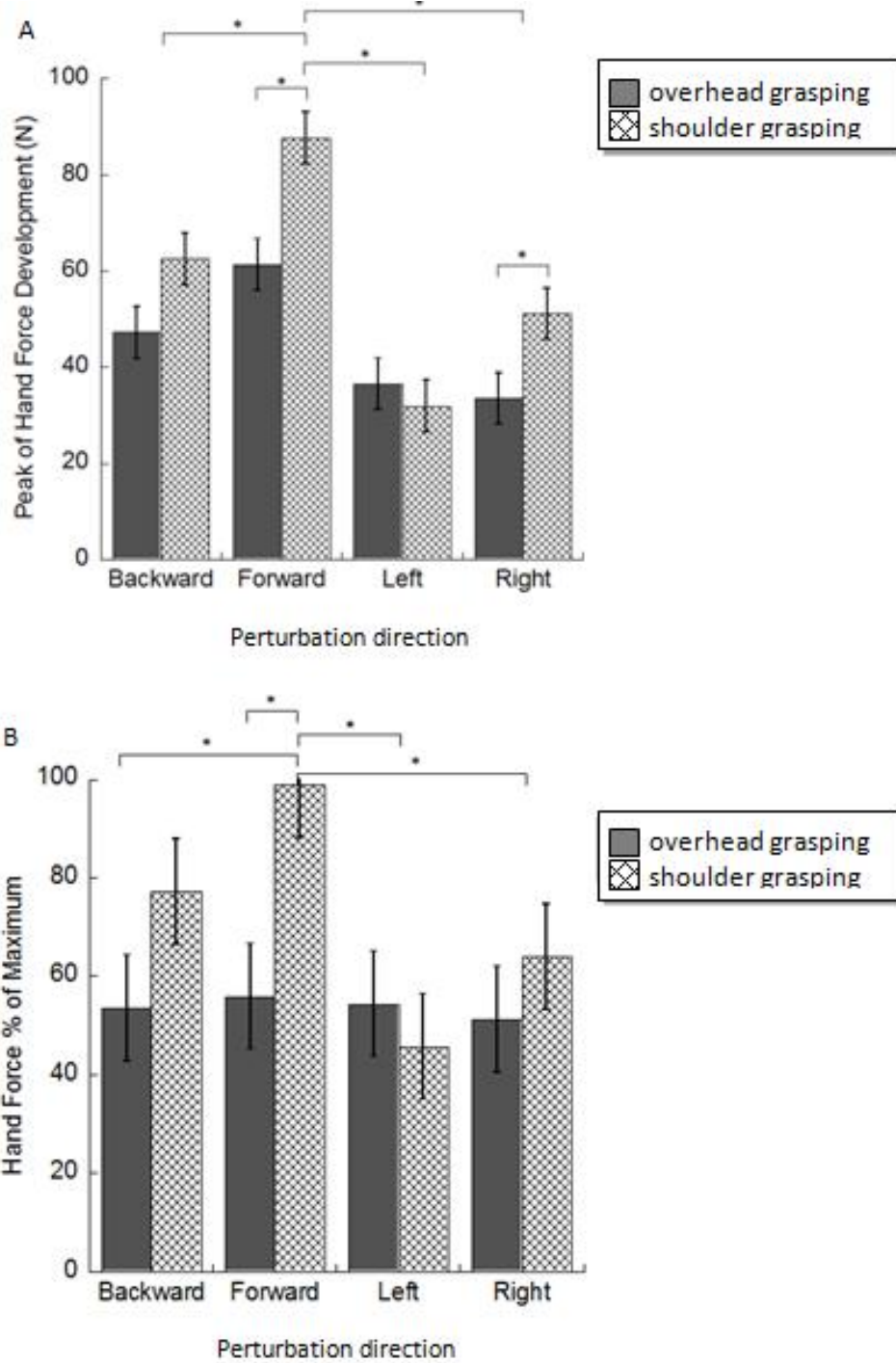


Figure 8: Effect of perturbation direction and hand configuration on (A) raw magnitudes of peak hand force and (B) peak hand force % of maximum. Average values and standard errors are shown. Asterisks show significant differences ($p < 0.05$). Forward perturbations while holding shoulder height handrail involved higher hand forces than all the other configurations.

Shoulder height handrail grasping

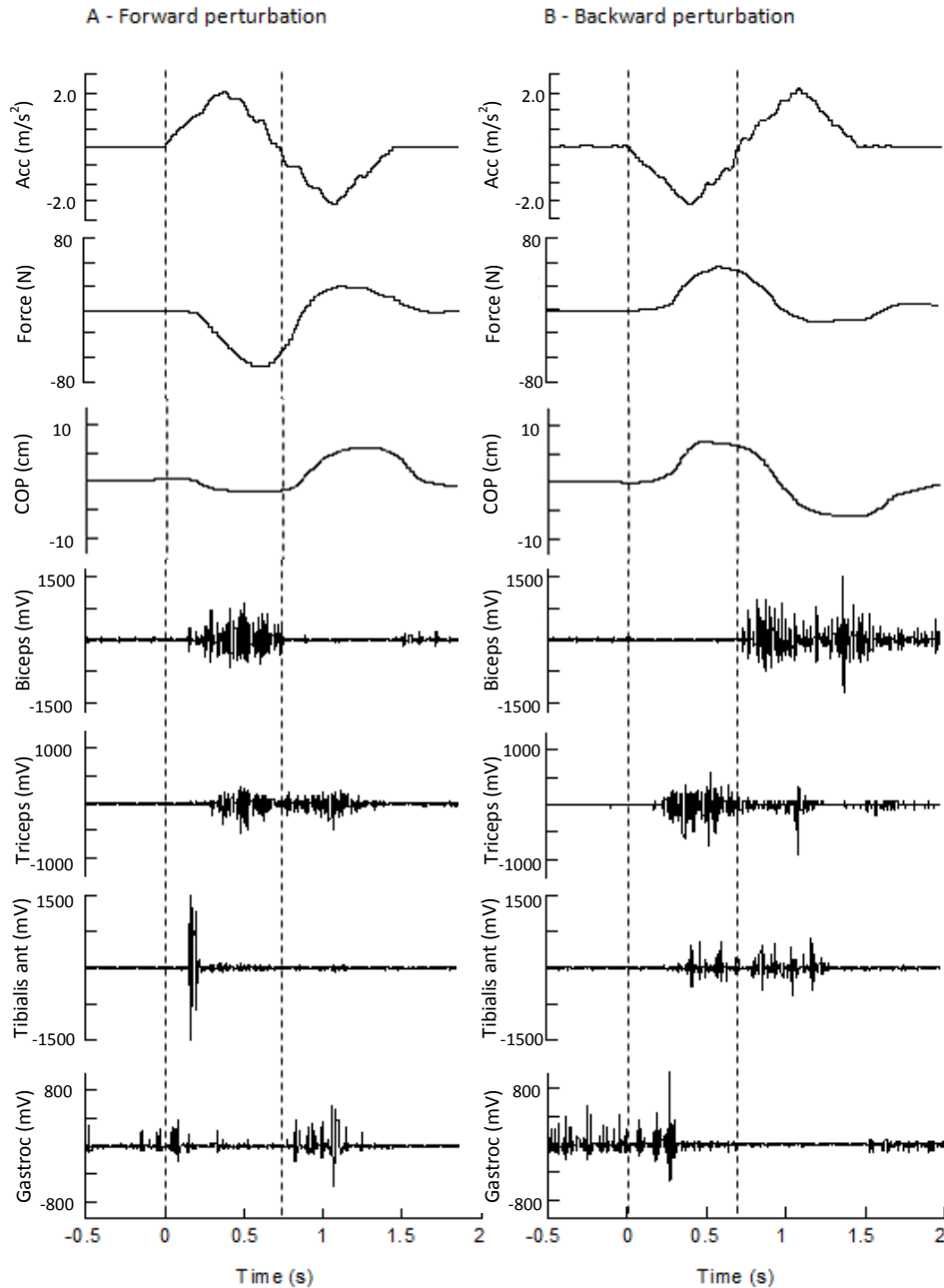


Figure 9: Example of EMG (biceps, triceps, tibialis anterior and gastrocnemius) obtained from 1 subject on a single trial during shoulder height grasping. Perturbation direction was set for (A) forward and (B) backward. The first dashed line represents the onset perturbation and the second the time where the acceleration was = 0 and the cart started to decelerate (see methods). Differences can be observed on hand force development, where higher hand forces are required during forward perturbation. These higher forces reflected higher EMG biceps activation. The higher forces generated by the hand, decreased the COP displacement to the backward direction. On the other hand, backward perturbations required more from the triceps and tibialis anterior muscles to help restoring balance.

Overhead handrail grasping

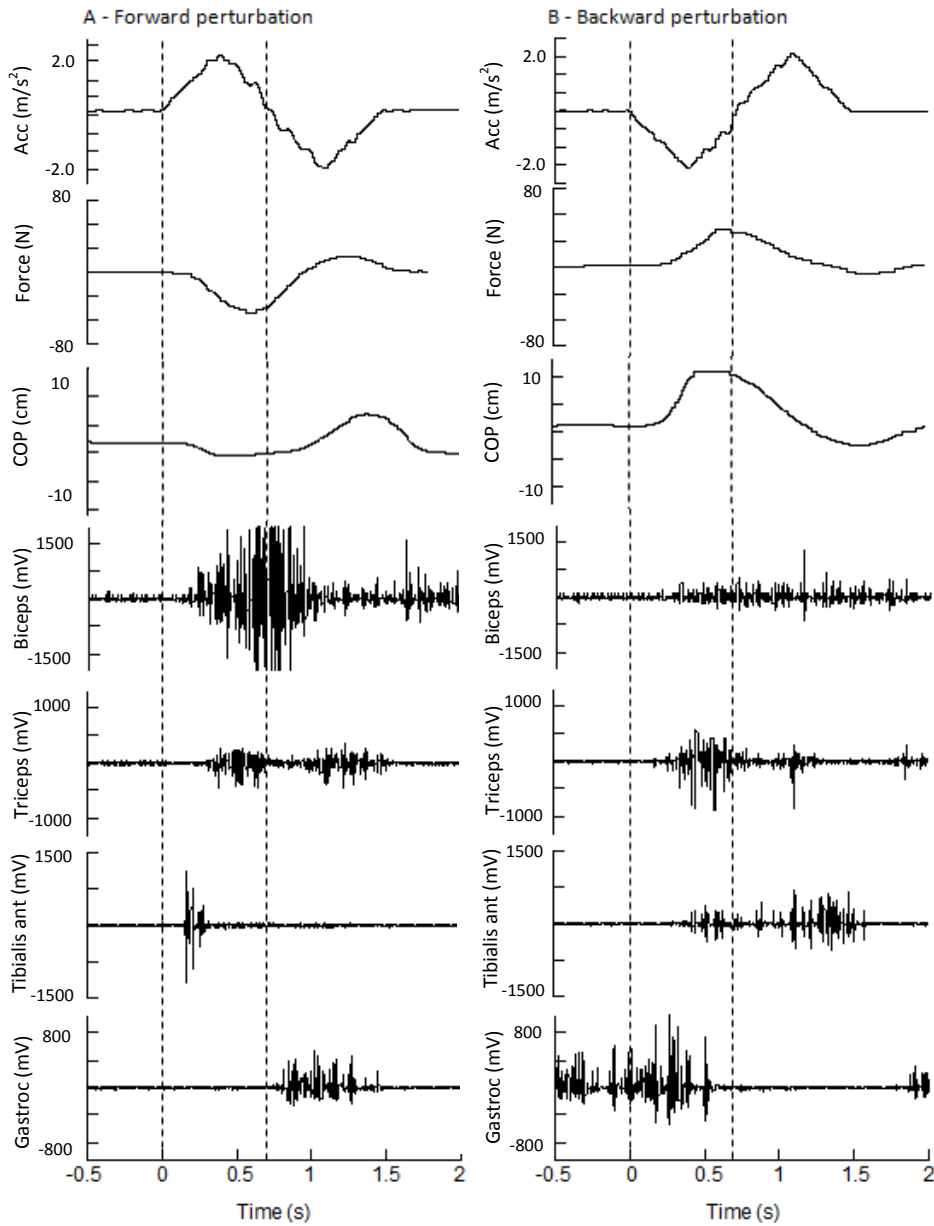


Figure 10: Example of EMG (biceps, triceps, tibialis anterior and gastrocnemius) obtained from 1 subject on a single trial during overhead grasping. Perturbation direction was set for (A) forward and (B) backward. The first dashed line represents the onset perturbation and the second the time where the acceleration was = 0 and the cart started to decelerate (see methods).

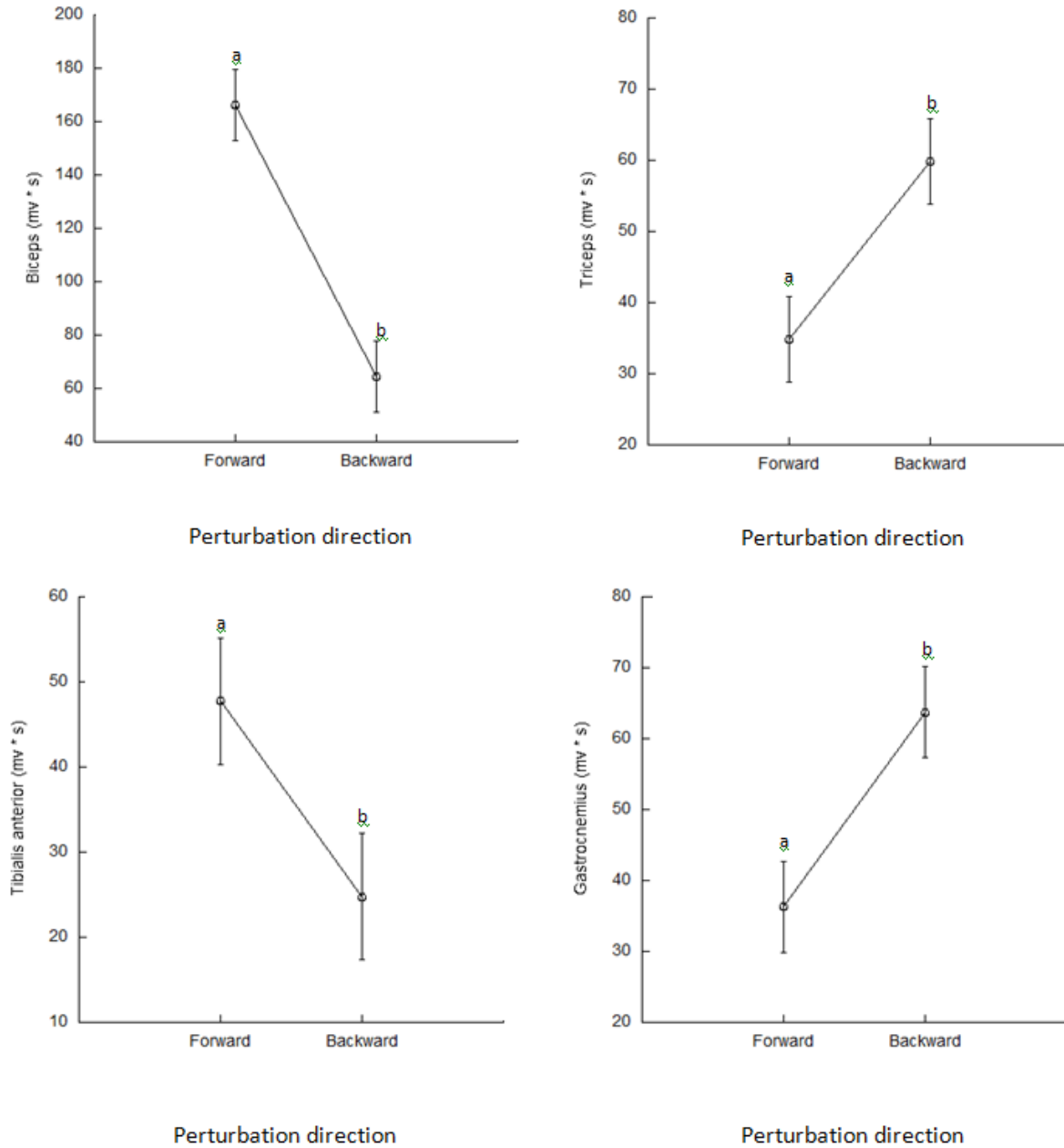


Figure 11: Muscle activity response from upper and lower limbs during forward and backward perturbation direction. The EMG signal calculated by integrating the muscle activity from the time of onset perturbation to the time of maximum perturbation velocity. The result shows that perturbation direction influences the muscle activity. Forward perturbation elicits the anterior muscles, while backward perturbations elicit more the posterior muscles to help bring the COP inside the base of support. Average values and standard errors (vertical bars) are shown. Levels not connected by same letter (i.e. *a* and *b*) are significantly different ($p < 0.05$).