Linear and Angular Head Accelerations in Daily Life

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(ABSTRACT)

The purpose of this study was to determine the linear accelerations, angular rates, and angular accelerations of the center of gravity of the head during daily activities and to determine the effect of angular terms on the linear accelerations. A total of 700 experiments were conducted with 18 subjects performing 13 different tasks. Resultant maxima were 93.6 m/s² for linear acceleration, 931.3 rad/s² for angular acceleration and 9.03 rad/s for angular rate. Comparisons by gender were statistically significant in 21.9% of cases. Qualitatively, subject effort appeared to be the most important factor.

Average error was strongly influenced by the type of motion in each event, ranging from -3.1% to 115.2% when converting from mouthpiece resultant accelerations to center of gravity acceleration. Error increases as angular rates and accelerations increase. Mouthpiece array accelerations are statistically significantly different than center of gravity accelerations in 86.3% of comparisons. Array designs from the literature are significantly different than center of gravity accelerations with equal frequency to the mouthpiece. Peak accelerations from Allen et al 1994 may require correction of up to 2G and 80% to obtain center of gravity accelerations. Angular terms must be accounted for even at daily activity levels.

Disclaimer

Mention of company names or products does not constitute endorsement in any way by the Virginia Polytechnic Institute and State University.

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For my family, without whom none of this would have been possible.

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Chapter 1

Introduction and Background

This chapter contains an overview of head accelerations in daily activities. The basic anatomy of the head is given along with research describing the location of the center of gravity of the head. The relevant research is summarized, highlighting major previous efforts to document daily activities.

The next two chapters are stand-alone papers describing the linear accelerations, angular accelerations, and angular rates found in this study (Chapter 2) and the effect of the angular terms on linear accelerations (Chapter 3).

1.1 INTRODUCTION

Brain injury is one of the longest studied areas of medicine, with trepanning – removal or boring of skull bones – dating back to at least 10,000 BC and concussion documented in both the Old Testament and in Hippocrates's Aphorisms (circa 415 BC). Concussions and other non-penetrating brain injuries such as aneurysms, despite millennia of awareness, have proven to be one of the hardest problems in medicine as they often occur with no observable physical injury to the brain matter (Shaw 2002). Brain injury is estimated to strike 1.5 million people annually in the US, with a loss of economic productivity of $56.3 billion in 1995 (BIAA 2002). It is only in the last 60 years that systematic theories have been put forth for how head acceleration could be the root cause of these types of injuries. Even now debate rages over whether linear or

angular acceleration is the primary cause of brain injury, or even whether these just correlate to injury (Shaw 2002, Zhang et al 2004). That said, linear and/or angular acceleration, or an index based on those, are the most often cited data for brain injury thresholds.

It has been recently noted, often by low-speed vehicle impact and amusement ride injury litigation, that there is little data regarding what constitutes “normally occurring” brain accelerations (Castro et al 1997, Exponent 2002). While there are extensive studies of head accelerations for vibration exposure and ergonomics, little has been published for impact-related peak accelerations in daily living. Currently, no data exists regarding angular accelerations resulting from everyday exposures and most linear acceleration studies have been limited to uni- or bi-axial analyses which make their comparison to injury thresholds difficult to impossible. At the present time there is no baseline for brain injury.

This study describes a two part analysis of the linear and rotational accelerations of the head in everyday life. The purpose of part I was to document complete six degree-of-freedom (DOF) data of the linear and rotational accelerations seen in a variety of daily living activities. The purpose of part II was to analyze the influence of rotational acceleration and velocity on the resultant acceleration vector and determine whether prior neglect of this data in previous research was justified.

1.2 BACKGROUND

1.2.1 Skull and Brain Anatomy

To understand head injury, one must understand the anatomy of the head. The head primarily consists of the skull, the brain, and the sensory organs of the face. In a mid-sized adult male, the head weighs approximately 10lb (4.54 kg) (Walker et al 1973). The skull is generally considered to consist of 22 bones – not counting the hyoid, teeth or the bones of the ear – of which only the lower mandible articulates. This is further divided into the 8 bones of the cranium, which serves as a helmet for the brain, and the 14 bones of the face (Gray 1918).

The brain, in a medical as opposed to embryonic scheme, consists of four major regions – the cerebral hemisphere, the diencephalon, the cerebellum, and the brain stem – enclosed within the cranium. The cerebrum is the largest portion of the brain, occupying the vault of the cranium. Consisting of the white matter, the gray matter, and the basal nuclei, this is regarded as the site of the intellect. Moving inferiorly and posteriorly, surrounded by the cerebrum is the diencephalon, which is primarily made up of the thalamus, hypothalamus, and epithalamus. This is a control and input center of the brain. Inferior to the diencephalon is the brain stem, which consists of the midbrain, pons, and medulla oblongata. The medulla is the inferior-most part of the brain and merges with the spinal cord at the level of the foramen magnum. This region controls automatic behavior. The final region is the cerebellum, which occupies the inferoposterior region of the skull. The cerebellum controls fine motor activity and may also control thought about motion. In an adult male, the brain weighs approximately 3.5 lb (1.59 kg) (Marieb 2001, Gray 1918). One school of thought regarding concussion and other head injury is

that effects tend to move from the superficial surface inward toward more central portions of the brain with progressing injury severity (Shaw 2002).

When establishing reference systems for the head, the most common external landmarks are the nasion, inion (external occipital protuberance), opisthion, glabella, external auditory meatus, and the optical orbit. The nasion is the central point of the frontonasal suture. It roughly corresponds to the depression at the root of the nose, between the eyebrows. The inion is the peak of the occipital protuberance of the occipital bone. The opisthion is the midpoint of the posterior edge of the foramen magnum. The glabella is the most prominent point in the mid-sagittal plane between the eyebrows (Gray 1918). Located slightly above, it sometimes replaces the nasion in reference plane definition. The external auditory meatus is the entrance of the ear and the orbit is the exterior portion of the eye socket. These are illustrated in Figure 1.

For purposes of reference planes, the two most commonly cited for cg location are the Frankfort horizontal plane and the nasion-inion plane. Both planes lie normal to the mid-sagittal plane and have the external auditory meatus at about the midpoint. The Frankfort horizontal plane is defined by a plane passing through the inferior rims of the orbits and the superior edges of the external auditory meatuses. The standard reference position of the skull places this plane horizontally. The nasion-opisthion plane, or plane of Beauvieux, is defined by a line between the nasion and opisthion. When held horizontally, this plane somewhat better corresponds to the animal skull models and places the cg in a more neutral position (Clauser et al 1969). The Frankfort line and plane of Beauvieux are shown in Figure 2.

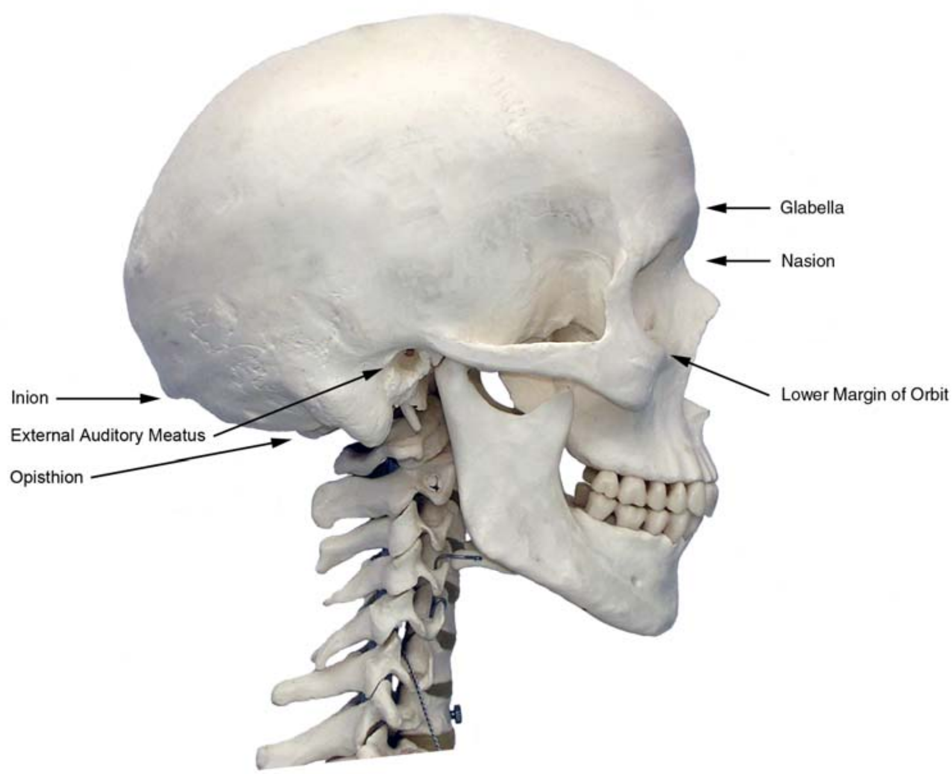

Figure 1. Skull landmarks.

1.2.2 Center of Gravity

The absolute motion of any point on a rigid body can be characterized if the acceleration of the center of gravity (cg) and relative motion of the point about the cg is known (Hibbeler 1978). Correspondingly, if one knows the location of the cg, an acceleration measurement taken at any other location can be solved for the cg acceleration (Chou and Sinha 1976). While the skull is certainly not a rigid body, measures can be taken to ensure that this approximation is useful for many analyses of the head and for accelerometry purposes is sufficient (Becker 1972).

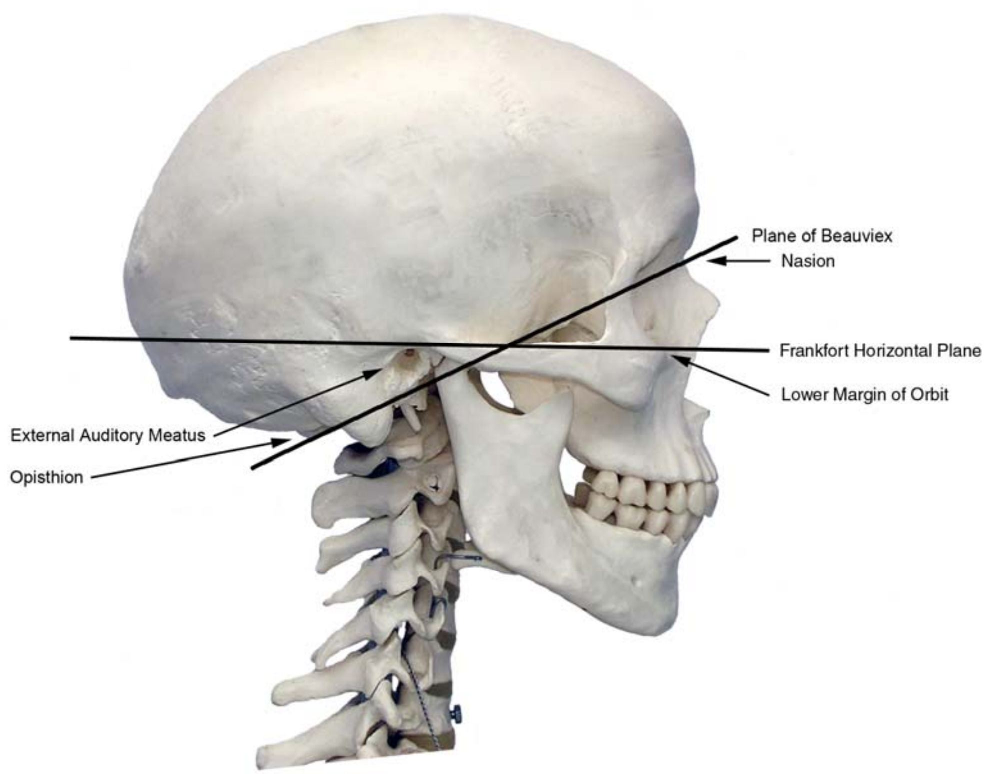

Figure 2. Planes of the skull.

The primary difficulty is in locating the cg of the head. The skull, as described previously, is an irregular ellipsoid composed of multiple bony plates, intracranial protrusions, and moving parts (lower jaw) that is devoid of obvious axes or reference planes (Gray 1918, Olivier 1978, Ricketts et al 1976). Researchers have typically referenced the Frankfort horizontal plane or the nasion-opisthion line when describing cg location, although human surrogate development has often referenced the vertex and rear of the skull. The cg has historically been found to lie in the mid-sagittal plane of the head, as the head exhibits bilateral symmetry.

Dempster, in a major human dimensions study, found the cg of the head alone to be located "…near the nasion-inion line at a point about 32 percent back from the nasion". He also describes it as a point in the sphenoid sinus, 4 mm beyond the antero-inferior margin of the sella turcica (Dempster 1955). Clauser, in a literature review for his own study, describes Dempster's point to be 43.3% of the distance from the joint center (presumably the foramen magnum) to the crown of the head. Clauser found the cg of the head to be located 11.15±0.21 cm below the vertex and 7.98±0.17 cm anterior to the back of the head based on 13 subjects, and provided regression equations based on anthropological measurements. The Naval Biodynamics Laboratory categorized small, medium, and large aviators between 1980 and 1990. The cg of small (3rd percentile) aviators was located 10.2 cm anterior to the back of the skull and 10.4 cm inferior to the vertex (Naval Biodynamics Laboratory 1988). The cg of medium (50th percentile) aviators was located 10.4 cm anterior to the back of the skull and 10.5 cm inferior to the vertex. The cg of large (95th percentile) aviators was located 10.5 cm anterior to the back of the skull and 10.6 cm inferior to the vertex. These results assume bilateral symmetry across the mid-sagittal plane.

Becker analyzed five cadaver heads and found the cg to be approximately 2 cm superior and 1.5 cm anterior to the auditory meatus, with the meatus as the origin and the anterior-posterior axis lying along the Frankfort plane. The margin of error was approximately 1cm in both directions. Using the same setup with 21 cadavers – 19 male and two female – Beier found the cg to be located 3.12±0.56 cm above the Frankfort plane and 0.83±0.25 cm anterior to the auditory meatus. The lateral offset was -0.05±0.13 cm (Beier et al 1979).

Walker studied twenty cadavers and found the cg both visually and radiographically to be located in an arc approximately centered at the antero-superior point of the ear. Beier describes this point to be 1.42±0.76 cm anterior and 2.41±1.03 cm superior to the auditory meatus in the Frankfort plane. This is shown in Figure 3.

Mertz analyzed the cgs of two cadavers of different size and developed a regression to locate the cg of living subjects (Mertz 1967). Hubbard calculated that for a 50^{th} percentile population, Mertz' cg location is 1.27 cm superior and 0.25 cm posterior to Walker's (Hubbard and McLeod 1974).

During development of Hybrid III human crash testing surrogate, the head inertia properties were primarily based on studies by Hubbard and McLeod. Hubbard and McLeod used the Walker and Mertz results as guidelines, and accepted Walker's results as adequate, as it was the largest study of head cg at that time. As such, Walker's data is the basis for current crash dummy cg location (Hubbard and McLeod 1974). Hubbard and McLeod's coordinate system was centered at the nasion, parallel to the Frankfort plane. The cg of the head, then, was 7.62 cm posterior and 1.27 cm inferior to the nasion. Tennant, describing the development of the ATD-502, which was an intermediate that would largely be carried over to the present Hybrid-III, places the cg as 4.3 in (10.9 cm) anterior to the rear of the head and 4.1 in (10.4 cm) inferior to the vertex, with the head in a Frankfort plane (Tennant et al 1974). These numbers are similar to the naval figures, although somewhat more inferior from the vertex. When using scaling factors to determine sizing for the male 95^{th} and female 5^{th} percentile dummies, Mertz found the distance in the infero-superior (vertical) direction from the occipital condyles to the cg of the head should be 2 in (5.1 cm) in the 95^{th} male and 1.8 in (4.6 cm) in the 5^{th} female,

versus 1.9 in (4.8 cm) in the 50^{th} percentile Hybrid III. Antero-posterior adjustments were not possible due to limitations of the scaling algorithm and shape differences in the populations.

Vital and Senegas deviated somewhat from the typical Frankfort plane reference and instead used the nasion-opisthion plane of Beauvieux. Studying six cadavers of mixed gender, head type and size, they found that the cg lies in the middle of the nasion-inion line, slightly behind the sella turcica and above and slightly in front of the external auditory meatus. They concluded the cg was located in a 1 cm^2 area "centered on the antero-superior implantation of the helix" or external ear (Vital and Senegas 1986). This location is very close to the cg found by Walker.

This study uses Walker's cg location because of its close proximity to a prominent external landmark and repeated validation in the literature. Subject head cg location will be approximated to be the antero-superior aspect of the helix of the ear. This point is shown on the skull in Figure 3.

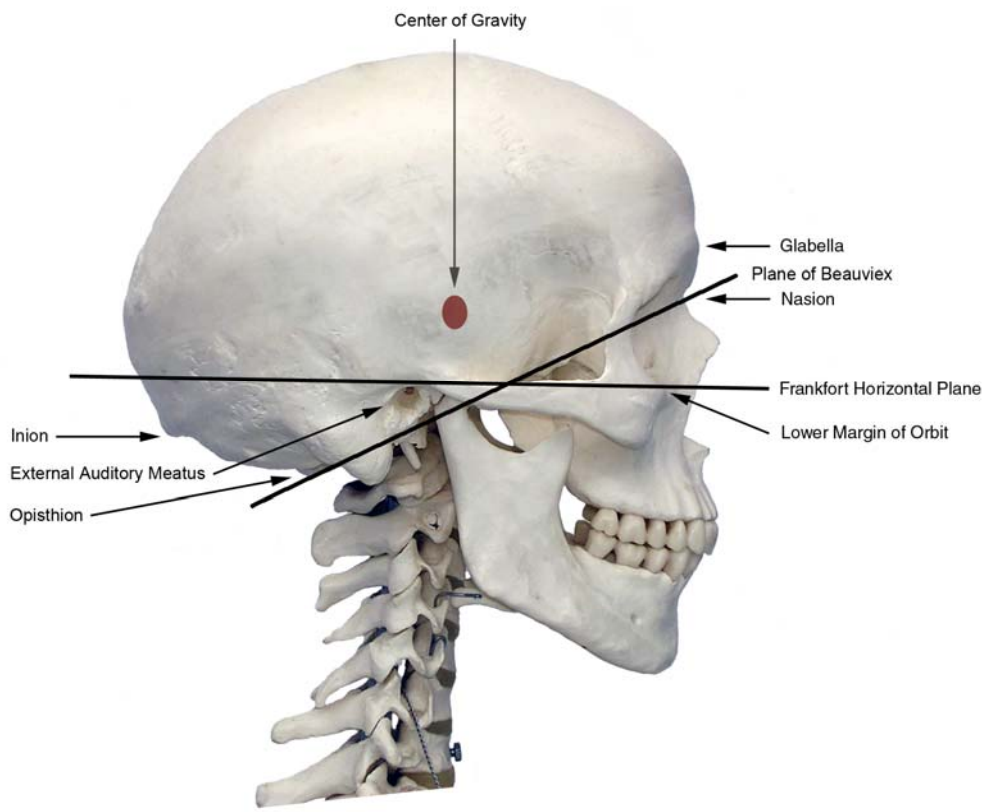

Figure 3. Center of gravity of head with reference landmarks. Center of gravity located 1.42±0.76 cm anterior and 2.41±1.03 cm superior to the auditory meatus in the Frankfort plane.

1.2.3 Daily Accelerations

Despite the volume of literature concerning impact accelerations of the head, especially in the sports or automotive safety fields, little has been written concerning the impacts and insults absorbed by the head during everyday life. The primary contributions are a study of daily perturbations by Allen and a test series conducted by Exponent as part of a study of rollercoaster safety for Six Flags, Inc.

Allen noted prior 'human tolerance' studies had produced a wide range of values, with reports of 2.9 and 5 g as non-injurious, and minimal symptoms at 4.8g and the like, often in the study of low-speed motor vehicle accidents. Boxers, however, repeatedly

tolerated 70 g blows with durations of 0.1 s. Seeking to identify "normal tolerance", especially in regards to whiplash, Allen instrumented eight subjects with a helmet-mounted triaxial accelerometer and exposed them to 12 different events from daily life. One subject was repeatedly tested in a 13th event. These events consisted of the following: look left; startle; shoulder check; stand up; head bob; sit down; sneeze; cough; crowd jostle; back slap; kick chair; hop off step; and plop in chair.

Discarding lateral accelerations as non-significant, Allen noted that "hopping off a step" or "plopping down into a chair" were capable of resultant accelerations of 8.1 and 10.1 g respectively. Allen concludes that the daily accelerations levels appear higher than those from no-damage vehicular accidents, although individual differences or pre-existing conditions may allow injuries at these levels. None of Allen's tests involved a direct impact to the head. King, in commentary on Allen, was somewhat critical of the magnitude of accelerations reported, and suggested that the helmet-mounting may be suspect – preferring a bite-plate measurement instead (Allen et al 1994).

Exponent, as part of a rollercoaster safety study, investigated both impact-type and g-induced loss of consciousness head injury potential. As part of the impact-type analysis, they analyzed daily accelerations in a manner similar to Allen, using seven test conditions, of which six did not involve direct impacts to the head. These conditions were the following: swing; obstacle course; sneeze; falling down; pillow strike; pogo stick; and spinning on a tire swing. Exponent instrumented nine subjects with wearable data-acquisition system and three groups of triaxial blocks, but did not report any angular accelerations. Unlike Allen they did report both duration and Head Injury Criteria (HIC) scores, which accounts for both magnitude and duration. Accelerations over 16 g and

HIC scores of 15.2 were generated in non-contact tests. Accelerations over 28 g and HIC scores of 22.6 were generated by a pillow blow to the head. Running an obstacle course, falling down on a mat, and the pillow strike all generated lateral accelerations over 2.5 g. Arndt, working from the same data, noted that rollercoaster rides could achieve HIC scores as high as 10 (Arndt et al 2004). The authors noted that a simple sneeze generated nearly 4 g in the x-direction (posterior to anterior) and that all test conditions would be considered benign by most people. They also noted that every day non-traumatic impacts such as bumping heads against tables or cupboards were neglected in their study. Neither Allen nor Exponent included the effect of gravity in their z-axis results (Exponent 2002). Exponent's and Allen's results are compared below in Figure 4.

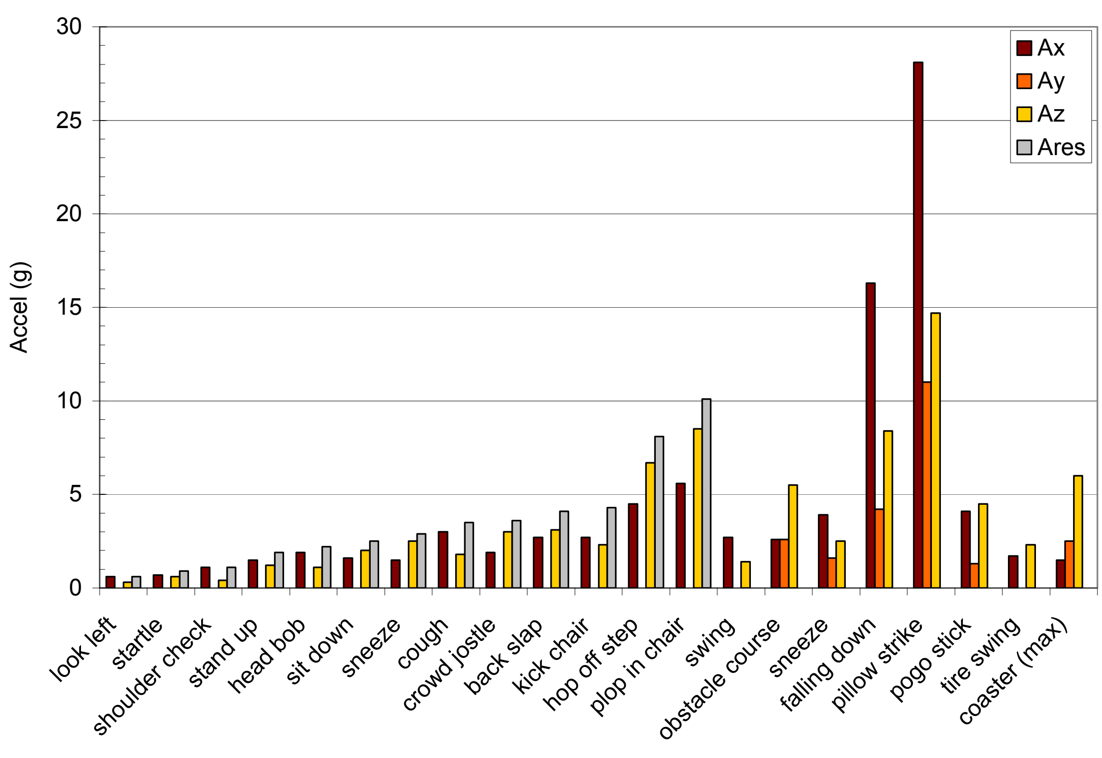

Figure 4. Results from Allen and Exponent.

Walking, running, and associated activities have received some attention as well. Kavanagh instrumented eight young and eight elderly subjects with triaxial accelerometers and recorded values of approximately 0.5 g in the vertical direction and 0.2 g in both the anterior-posterior and medial-lateral directions for both populations (Kavanagh et al 2004). Woodman and Menz both agreed with those numbers, but Woodman found an anterior-posterior acceleration of 0.5 g. Angular accelerations were also recorded by Woodman, with approximately 40 rad/s^2 in pitch and 10 rad/s^2 in both roll and yaw (Woodman and Griffin 1996, Menz et al 2002). Mahar compared running and inline skating on a treadmill and found head accelerations in skating were similar to those of walking, while 6 mph running produced peak accelerations approximately double those of walking or skating (Mahar et al 1997). Mercer reported 1.5±0.5 g and 1.7±0.4 g for rested and fatigued 3.8 m/s (8.5 mph) running, respectively (Mercer et al 2003). Standing in an elevator has also been investigated. Yung found that elevators averaged 0.1-0.2 g during normal operation and up to 0.7 g during emergency stops, in the vertical direction. These numbers do not include a gravity component (Yung and Gowda 2002).

Fast and Sosner investigated accelerations levels involved in rolling wheelchairs into curbs. Fast found that non-contact accelerations ranged as high as 1.9 g in the anterior-posterior direction, 1.7 g laterally, and 2.1 g vertically for restrained subjects (Fast et al 1997). Sosner found non-contact accelerations of 1.4 g in the anterior-posterior direction, 0.7 g laterally, and 1.1 g vertically for unrestrained subjects (Sosner et al 1997).

Some research has also been reported concerning low-speed automobile impacts. Szabo conducted ten tests at an impact speed of approximately 14 km/h (8.7 mph) generating an average of 6.5 g in the struck vehicle. Szabo instrumented five subjects, four male and one female, with a nine-accelerometer head array, although no angular or lateral linear accelerations results were reported. Resultant linear head accelerations of up to 17.2 g were reported, with maximum directional values of 14.8 g in the anterior-posterior direction, and 17.2 g in the vertical direction. No injuries were sustained by any subject during testing (Szabo and Welcher 1996). Castro conducted a similar study using nineteen subjects, fourteen male and five female, comparing low-speed vehicle crashes to bumper car impacts. The auto impacts occurred at an average speed of 11.4 km/h (7.1 mph) with an average acceleration of 2.7 g versus a speed of 9.9 km/h (6.2 mph) and acceleration of 2.2 g in the bumper cars. The authors noted five cases of minor injury and one persistent case of minor neck injury in the automobile series versus none in the bumper cars, despite similar acceleration profiles. Most injuries were located in the cervical spine. The authors were uncertain whether the limit of harmlessness lied between the auto and bumper car series, or whether psychological factors were also at play (Castro et al 1997). The BIAA echoed findings that injury had been reported in vehicular rear-impacts as low as 2-3 g (BIAA 2002).

Bumper cars are not the only amusement rides studied. Beyond the Exponent report described previously, other authors have focused on roller coasters and other amusement rides. Smith and Meaney investigated three roller coasters for non-contact head accelerations, reporting maximums of 4.2 g for lateral, 5.4 g for anterior-posterior, and over 5 g for vertical linear acceleration and angular accelerations of 502 rad/s^2. The

authors noted that those acceleration levels were nine times too low for vein tearing and 18 times too low for diffuse axonal injury in the brain. They were also much lower than non-concussive results from amateur boxing (Smith and Meaney 2002). The Brain Injury Association of America reported sustained acceleration results for a variety of flat rides as well as an average flume-type water ride and an average launched roller coaster. No ride generated more than 2.5 g in the anterior-posterior direction, 2.0 g in the lateral direction, or 6.0 g in the vertical direction, inclusive of gravity (BIAA 2002).

Sports have also received attention, especially impacts in American football, soccer, and boxing. Atha investigated a professional boxer and found that a normal punch was capable inducing 53 g in the target (Atha et al 1985). Pincemaille found similar numbers using volunteer amateur boxers, documenting linear accelerations ranging from 18-79 g at minimum 3 ms durations. Angular accelerations regularly exceeded 3500 rad/s^2 (Pincemaille et al 1989). Duma investigated collegiate football and in a study of 3312 impacts, found linear accelerations averaging 32±25 g and angular accelerations averaging 905±1075 rad/s^2 about the x-axis and 2020±2042 rad/s^2 about the y-axis. They documented a concussion at 81 g, 5600 rad/s^2 x-axis rotation and 5590 rad/s^2 y-axis rotation, and a neck injury at 86 g, 4132 rad/s^2 x-axis rotation and 6638 rad/s^2 y-axis rotation (Duma et al 2005). Pellman, in a five year study of concussive impacts in the NFL, found that peak head accelerations in players sustaining concussions averaged 98±28 g, and that peak head accelerations in players not receiving concussions averaged 60±24 g. These numbers agree well with Duma and boxing reports (Pellman et al 2003). Naunheim studied heading in soccer, at ball speeds of 9 and 12 m/s (20 and 26 mph), documenting resulting linear head accelerations of 16.1±1.9 g and angular

accelerations of 1302±324 rad/s^2 at 9 m/s and linear accelerations of 20.3±2.8 g and angular accelerations of 1457±297 rad/s^2 at 12 m/s. These results were found to be well below expected tolerance for a single blow, but the effect of repeated acceleration at this level was unknown (Naunheim et al 2003).

Despite widespread popularity, neither contact sport nor amusement ride accelerations are events expected in daily living. Nor can it be said that even low-speed motor vehicle collisions are seen on a daily basis by the general public. Only a small percentage of the public can expect to deal with curbs while in a wheelchair or with emergency elevator stops. What remains are the studies of Allen and Exponent, and the data regarding running and walking. These studies have historically neglected to report angular acceleration, even when recorded, and have tended to absorb their acceleration histories into other variables. There are still holes in our knowledge of y-axis linear acceleration and angular accelerations in all three axes. There is much work yet to be done to further our knowledge of accelerations seen in everyday life.

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Chapter 2

Linear and Angular Head Accelerations in Daily Activities

ABSTRACT

OBJECTIVE: The goal of this study was to determine the linear head accelerations and for the first time, the angular rates and accelerations in daily activities.

STUDY DESIGN: Subjects were instrumented with a mouthpiece consisting of three orthogonal linear accelerometers and three angular rate sensors. Eighteen subjects completed thirteen activities for 700 samples. Results were compared by gender and against the previous literature.

RESULTS: Resultant maxima were 93.6 m/s^2 for linear acceleration, 931.3 rad/s^2 for angular acceleration, and 9.03 rad/s for angular rate. Maximum Vertical Leap was the most forceful activity with maximums in linear and angular acceleration and large angular rates. Gender differences were significant in 21.9% of comparisons. Results agreed well with previous walking and running literature. Head Bob and Look Left were higher than in previous literature and Chair Plop and Stair Jump were lower. Qualitatively, effort difference between subjects appeared to have a larger effect than array placement, gender, or size differences.

CONCLUSIONS: For daily activities, maximum accelerations are less than 10G, 1000 rad/s^2 and 10 rad/s. Differences between male and female responses were caused as much by individual effort differences as gender or size differences.

2.1 INTRODUCTION.

The medical and engineering literature contains many studies of single or multiple "everyday life" activities using volunteers (Allen et al 1994, Arndt et al 2004, Exponent 2002, Kavanagh et al 2004, Mahar et al1997, Mercer et al 2003, Smith and Meaney 2002, Woodman 1996). The majority of these studies concentrate on linear accelerations and either neglect or neglect to report the angular components – angular velocity and angular acceleration. Previous studies have also largely ignored gender and possible differences in responses between men and women.

The purpose of this study was to determine the angular components of acceleration and possible gender differences in a variety of everyday activities

2.2 METHODOLOGY

2.2.1 Experimental Test Setup

A biteplate system was used to evaluate nine head acceleration components – three linear acceleration, three angular acceleration, three angular velocity – in 18 volunteer subjects. The biteplate consisted of a custom aluminum plate screwed to the drilled flange of a stainless steel dental impression tray (vendor). Three linear accelerometers (Endevco 7596A, 30 G, San Juan Capistrano, CA) with a resolution of 0.008 G were mounted on a 1 inch (2.54 cm) Delrin block in an orthogonal configuration, which was then screwed to the bottom of the mouthpiece. Three magneto-hydrodynamic angular rate sensors (ATA Sensors ARS-06S, Albuquerque, NM) in an orthogonal array were also screwed to the aluminum plate. This is shown in Figure 5. Biteplate stability relative to head was validated by high-speed video. The accelerometers were mounted such that there was a common point of measure in the middle of the triaxial block. This point was aligned with the midline of the head such that the y-axis offset from the center of gravity of the head was zero. Data was oriented according to the SAE dummy coordinate system given in SAE J1733. The completed biteplate mounted on a medium-sized impression tray (Harry J Bosworth Co #2-4 SS Solid Upper Tray, Skokie, IL) weighed 240 g (0.529 lb).

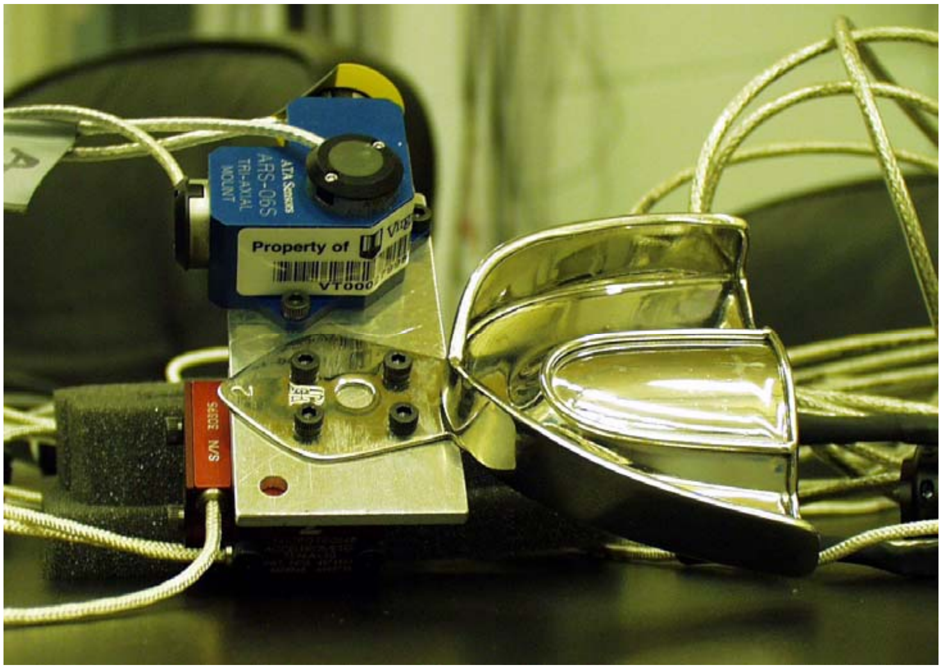

Figure 5. Assembled biteplate system. Triaxial block mounted along midline of head on bottom. Triaxial angular rate sensor array mounted offset on top of block. Plate attached to impression tray flange with screws.

Each subject selected an individual impression tray from a choice of three sizes. A custom plastic "boil and bite" mouthpiece was then made for each subject. Each subject was photographed from a side view while wearing the mouthpiece. The photographs were analyzed to measure the relative angle between the mouthpiece and the Frankfort plane, and offsets from the center of gravity were calculated. The center of gravity was taken as the antero-superior point of the helix of the outer ear. Positive in the x-direction was towards the front of the head along the Frankfort line. An example of this photograph is shown in Figure 6. Subjects were measured for height and weight in bare feet or wearing socks. Height was measured using a tape measure. Weight was measured on a digital scale (Conair WW17, East Windsor, NJ) All subjects signed a consent form and all test procedures were reviewed and approved by the Virginia Tech Institutional Review Board.

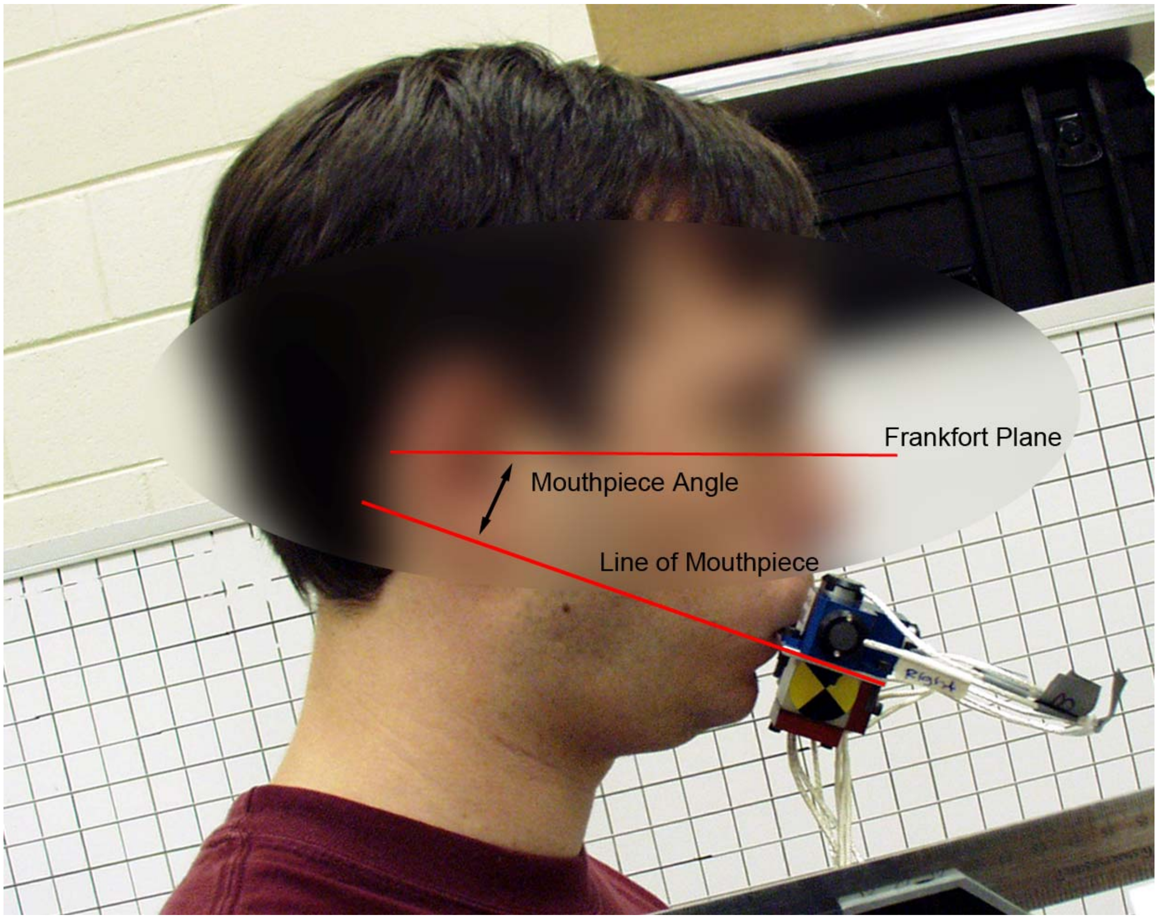

Figure 6: Example photograph showing Frankfort Plane, the line of the mouthpiece, and the correction angle between.

Subjects also wore triaxial accelerometers at the L5 and T1 levels of the spine. These were attached with double-sided tape, wrapped in a foam pad, and secured using self-adhesive medical bandage at the T1 level and Velcro-closed lumbar supports at the L5 level. Data from these sensors is part of another study and is not included here.

Subjects were asked to perform thirteen activities plus a baseline activity to determine channel outputs while standing with the head held in the Frankfort position. Each activity was performed three times and the results were averaged to provide an average response per subject per activity. The activities are illustrated in Table 1. With eighteen subjects, thirteen activities and three repetitions per activity, a total of 702

samples were recorded. As two repetitions were lost, one each from subjects 12 and 15, the results from 700 recordings are presented here.

Table 1: Activity list

Number	Event	Description
0	Baseline	Stand with Frankfort plane horizontal in order to zero the accelerometers
1	Chair Stand	Stand quickly from wooden chair
2	Normal Sit	Sit normally into wooden chair
3	Plop Sit	Sit quickly into wooden chair
4	Head Nod	Quickly nod head down and back up
5	Look Left	Quickly look left and return to center
6	Walk	Walk at 3 mph on treadmill
7	Run	Run at 6mph on treadmill
8	Run in Place	Run quickly in place
9	Jumping Jacks	Perform jumping jacks
10	Maximum Vertical Leap	Jump as high as possible, land on both feet
11	Stair Jump	Jump from one-stair height (~8")
12	Throw Ball	Throw 29 g (0.064 lb) foam ball at wall
13	Jump Shot	Jump-shoot basketball at target

Each subject was given a countdown before beginning the activity and was told to start at “Go”. This set the zero time for the data acquisition. Data was recorded for five seconds post-trigger and one second pre-trigger. Data from the accelerometers and rate sensors were recorded at a sampling frequency of 2,000 Hz with an Analog-to-Digital conversion resolution of 16 bits using an Iotech Wavebook with WBK16 strain gage modules (Iotech WBK16, Cleveland, OH). All linear accelerometer channels were filtered to CFC 180 with all rate sensor channels filtered to CFC 18. CFC filters were digital, 4-pole phaseless Butterworth filters.

Data was post-processed in MATLAB (The MathWorks MATLAB 7.0.1, Natick, MA) to correct for mouthpiece-Frankfort angle, to calculate angular accelerations, to calculate resultant accelerations and velocities, and to calculate linear accelerations at the center of gravity. All subjects were assumed to have zero Y-axis offset. Angular accelerations were calculated from the recorded angular rates using the symmetrical five-

point algorithm shown below. This equation was chosen because it reduces the error term to $O(dx^4)$

$$f'(x_n) = \frac{-f(x_{n-2}) + 8f(x_{n-1}) - 8f(x_{n+1}) + f(x_{n+2})}{12dx} \qquad \text{(eq 1)}$$

Linear accelerations at the cg were calculated based on mouth accelerometer traces based on the transformation below (Naumheim et al 2003). As Y-axis offsets are assumed zero, Y-based terms have been removed from the equations. X and Z are locations of the triaxial block center relative to the center of gravity.

$$a_x^{cm} = a_x^{triax} - (\alpha_y * Z) - \left[(\omega_x * \omega_z * Z) - (\omega_y^2 + \omega_z^2) * X\right] \qquad \text{(eq 2)}$$

$$a_y^{cm} = a_y^{triax} - \left[(\alpha_z * X) - (\alpha_x * Z)\right] - \left[(\omega_y * \omega_z * Z) + (\omega_y * \omega_x * X)\right] \qquad \text{(eq 3)}$$

$$a_z^{cm} = a_z^{triax} - (-\alpha_y * X) - \left[(\omega_x * \omega_z * X) - (\omega_x^2 + \omega_y^2) * Z\right] \qquad \text{(eq 4)}$$

2.2.2 Volunteer Subject Data

A total of 18 healthy subjects (9 male and 9 female) with an average age of 22.1 ± 2.9 years and range of 19-32 years, an average height of 66.9 ± 3.8 in, and an average weight of 157.7 ± 32.2 lb volunteered for this study. All subjects read and signed an informed consent form and had the opportunity to withdraw at any time from any or all parts of the procedure. All subjects completed all activities. Subjects were chosen for their approximate correspondence to one of four anthropometric targets – Hybrid-III 5th female, Hybrid-III 50th male, Hybrid-III 95th male, and 50th female target based on a combination of anthropometric studies (NASA 1978, Lenard and Welsh 2001, Huh and Bolch 2003, Welsh and Lenard 2001). Four subjects fit the small female category

corresponding to the Hybrid-III 5th female dummy, five subjects fit the mid-female category corresponding to a 50th percentile female, five subjects fit the mid-male category corresponding to a Hybrid-III 50th male dummy, and four subjects fit the large male category corresponding to a Hybrid-III 95th male dummy.

Mid-male and mid-female subjects closely approximated their corresponding anthropometric test devices (ATDs). The small female subjects were significantly taller and heavier than the 5th female ATD and large male subjects were significantly shorter and lighter than the 95th male ATD based on two-tailed z-tests. Despite this, both small females and large males were significantly different in mean height and weight based on a single-tailed Student's t-test than their 50th percentile counterparts. This is shown in Figure 7 and Table 2.

Table 2: Volunteer anthropometrics by category

Category	Height (in)	Weight (lb)	Target Height (in)	Target Weight (lb)	z-value (height)	z-value (weight)
SF	62.1 ± 1.8	118.3 ± 5.4	60	110	*0.016**	*0.002**
MF	65.2 ± 1.7	140.8 ± 6.9	64	140	0.122	0.786
MM	68.7 ± 1.0	168.6 ± 6.4	69	170	0.431	0.627
LM	71.5 ± 2.2	204.7 ± 6.7	74	223	*0.025**	*0.000**

* Significant difference between mean and target if z < 0.05.

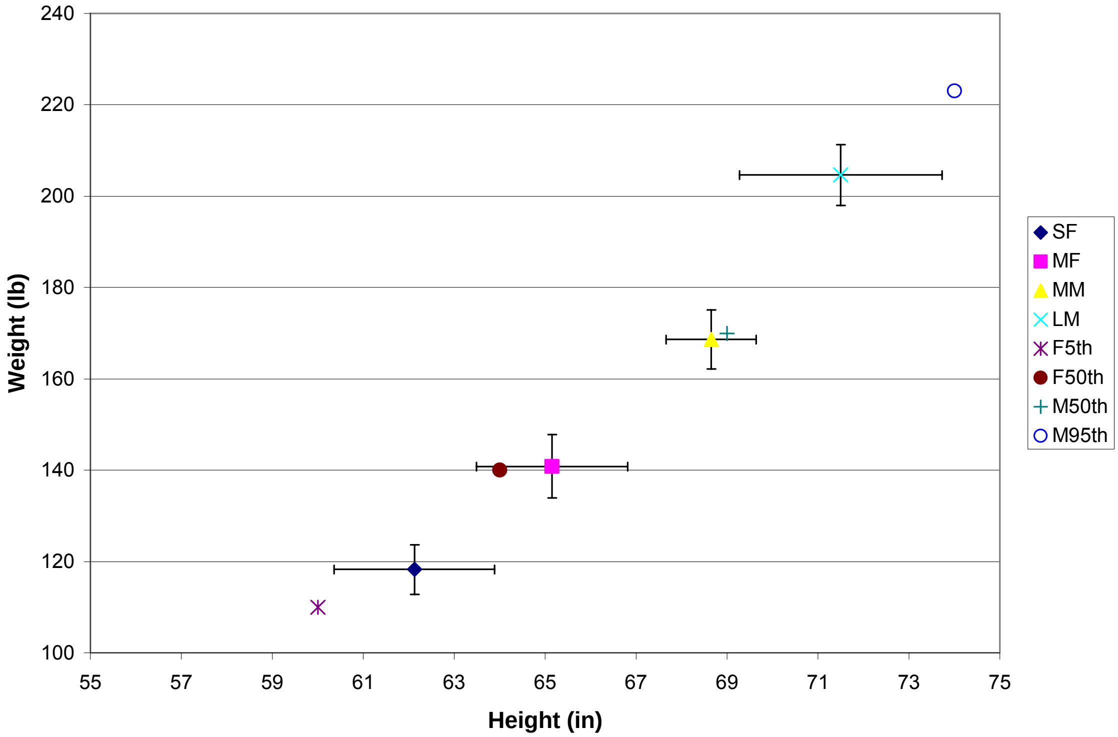

Figure 7: Volunteer categories vs target sizes. All data is mean ± standard deviation.

2.2.3 Data Analysis

The maximum positive and maximum negative accelerations from each channel plus resultants, angular acceleration, and cg acceleration were recorded from each sample trace. The three samples from each activity were averaged into a subject average for that activity. From this data category-specific, gender-specific, and overall summaries were calculated for each activity. To calculate inter-group differences, a Student's t-test was used, as sample sizes were less than the 30 required for a normalized p-test. If inter-category variance was not significantly different (<0.05) based on a f-test, then a two-tailed, "two test equal variance" t-test was performed. If variances were significantly different, then a two-tailed, "two test unequal variance" t-test was performed.

2.3 RESULTS

2.3.1 Average and Maximum Responses

No subjects reported any injuries of any kind. The results presented should be considered at a non-injurious level for a young adult population.

An overall average of the three-test average of peak linear acceleration data is shown below in Table 3 and Figure 8. The overall average of peaks is the average of the three-repetition average of the peak acceleration of each event. Gravity is not included in these results. With the head held with the Frankfort plane horizontal, gravity would add 9.81 m/s^2 in the +Z direction. The predominant peak acceleration directions are in +X (forward) and –Z (up), with peak x-axis accelerations usually the largest. In no events do y-axis accelerations reach the level of 1 g. Despite this, y-axis accelerations are dominant in Look Left. Maximum Vertical Leap is the most forceful activity with a resultant acceleration of 39.4 m/s^2 and peak scores in all three axes.

These trends are echoed by the event maximums. The overall event maximums are the maximum values for each category in each event across all repetitions by all volunteers. It is essentially a 'peak of peaks'. The maximums are much larger than the average of peaks – often approximately three times as large. The same general trends are followed with Maximum Vertical Leap the most forceful activity and Look Left the sole event dominated by y-axis acceleration. Stair Jump demonstrates slightly larger maximum accelerations than Running in Place and Jumping Jacks despite being somewhat smaller on average. This can be seen in Table 4 and Figure 9. The traces for the maximum event, subject 11's Maximum Vertical Leap, are shown in Figure 10.

Table 3. Overall average peak linear accelerations per event (m/s²)

Event		CG X	CG Y	CG Z	CG Res
1	Ave +	5.25	1.37	6.07	8.31
	Ave -	-1.98	-1.12	-6.56	
2	Ave +	5.08	1.20	3.61	7.74
	Ave -	-1.87	-1.29	-4.92	
3	Ave +	10.15	2.80	6.42	19.68
	Ave -	-4.35	-2.21	-16.51	
4	Ave +	4.44	0.93	4.35	12.45
	Ave -	-11.65	-1.06	-1.87	
5	Ave +	3.44	5.28	0.82	6.22
	Ave -	-2.85	-3.52	-1.25	
6	Ave +	3.30	2.16	3.18	6.40
	Ave -	-1.14	-1.77	-5.58	
7	Ave +	7.02	3.04	10.64	15.69
	Ave -	-3.72	-3.15	-14.46	
8	Ave +	12.36	4.04	12.62	30.10
	Ave -	-3.50	-3.58	-28.15	
9	Ave +	12.26	4.13	12.72	30.60
	Ave -	-3.79	-3.62	-28.80	
10	Ave +	20.32	6.89	21.30	39.38
	Ave -	-10.17	-6.75	-34.23	
11	Ave +	11.76	3.21	10.95	28.67
	Ave -	-5.86	-3.29	-26.70	
12	Ave +	6.54	5.55	4.39	10.06
	Ave -	-4.56	-3.85	-3.94	
13	Ave +	12.32	4.67	12.43	21.15
	Ave -	-2.18	-2.71	-17.41	

Table 4. Overall maximum linear accelerations per event (m/s²)

Event		CG X	CG Y	CG Z	CG Res
1	Max +	8.94	4.30	13.71	14.41
	Max -	-7.97	-2.10	-13.62	
2	Max +	8.05	3.10	9.16	15.13
	Max -	-12.95	-2.61	-12.22	
3	Max +	23.53	9.55	11.78	39.35
	Max -	-15.11	-7.70	-37.79	
4	Max +	10.83	2.01	13.54	27.41
	Max -	-25.65	-3.06	-4.50	
5	Max +	6.04	9.77	3.23	10.48
	Max -	-8.32	-7.33	-2.85	
6	Max +	6.69	4.14	4.99	10.05
	Max -	-3.51	-3.41	-9.17	
7	Max +	13.87	6.60	13.58	25.82
	Max -	-19.17	-8.29	-22.03	
8	Max +	29.83	9.09	20.79	40.90
	Max -	-7.69	-11.76	-32.69	
9	Max +	22.53	15.50	23.57	43.38
	Max -	-15.66	-14.71	-40.56	
10	Max +	57.62	28.13	35.63	93.62
	Max -	-52.84	-40.51	-87.21	
11	Max +	30.70	10.03	19.65	49.28
	Max -	-16.78	-17.50	-47.02	
12	Max +	12.81	16.21	11.18	19.30
	Max -	-18.22	-7.99	-10.29	
13	Max +	18.92	8.68	22.26	37.22
	Max -	-8.95	-4.53	-36.71	

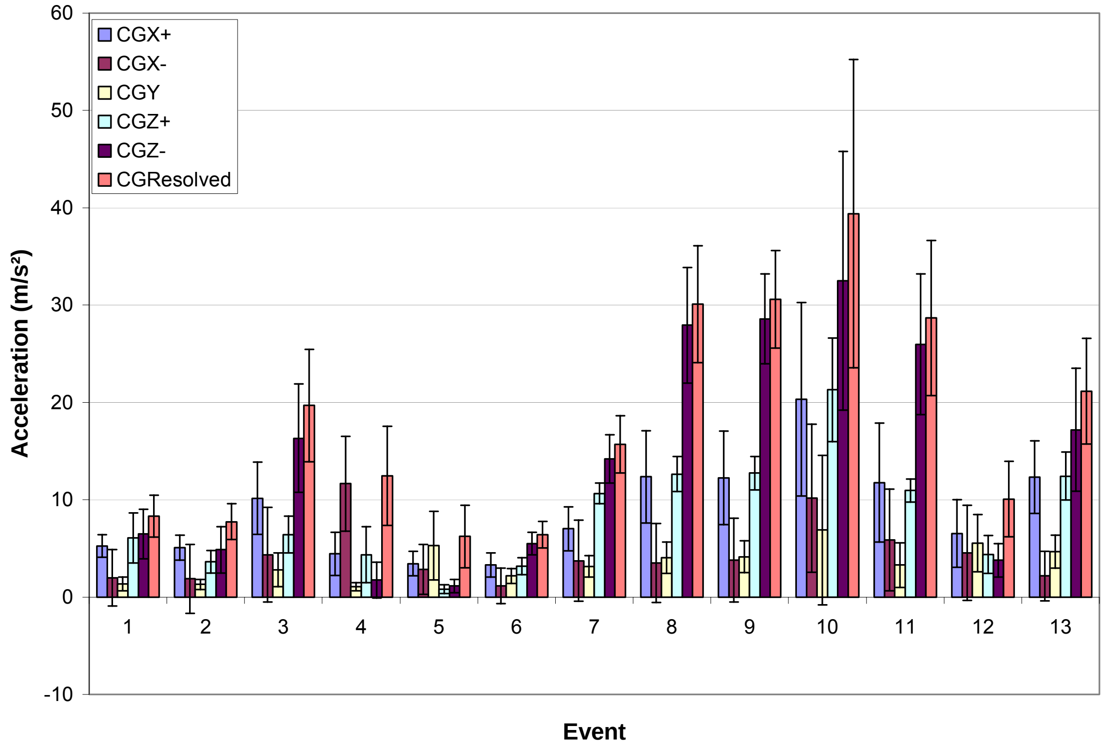

Figure 8. Average peak cg linear accelerations per event. Results are mean ± standard deviation.

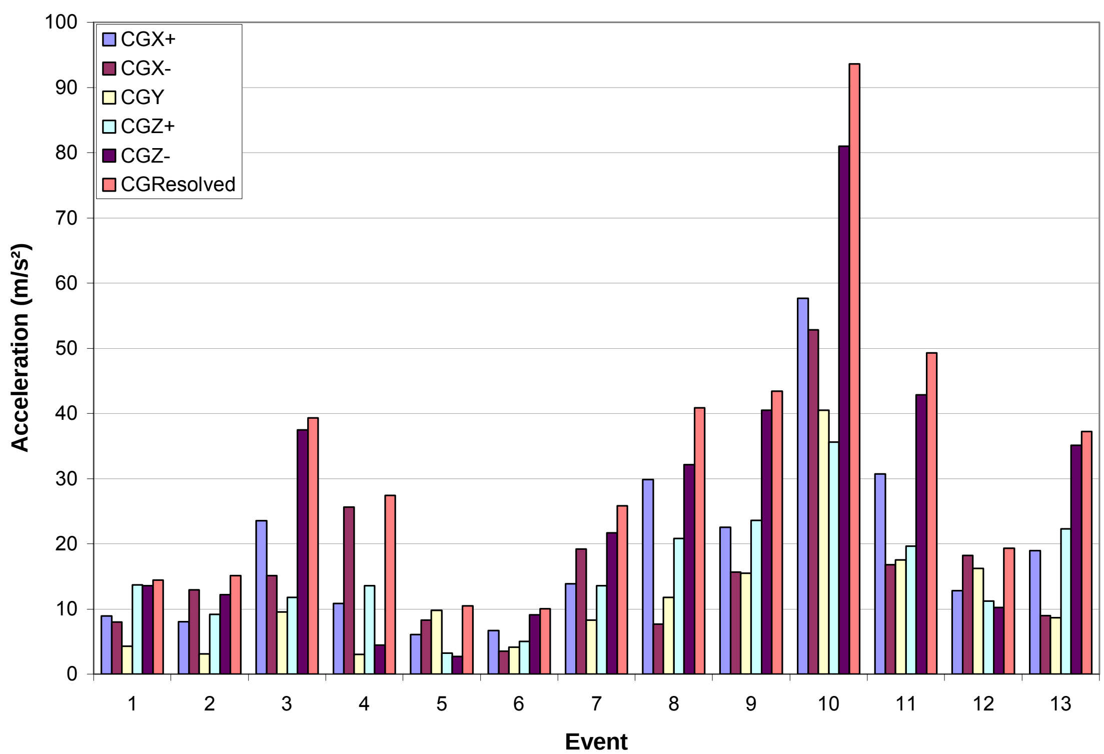

Figure 9. Maximum cg linear accelerations per event.

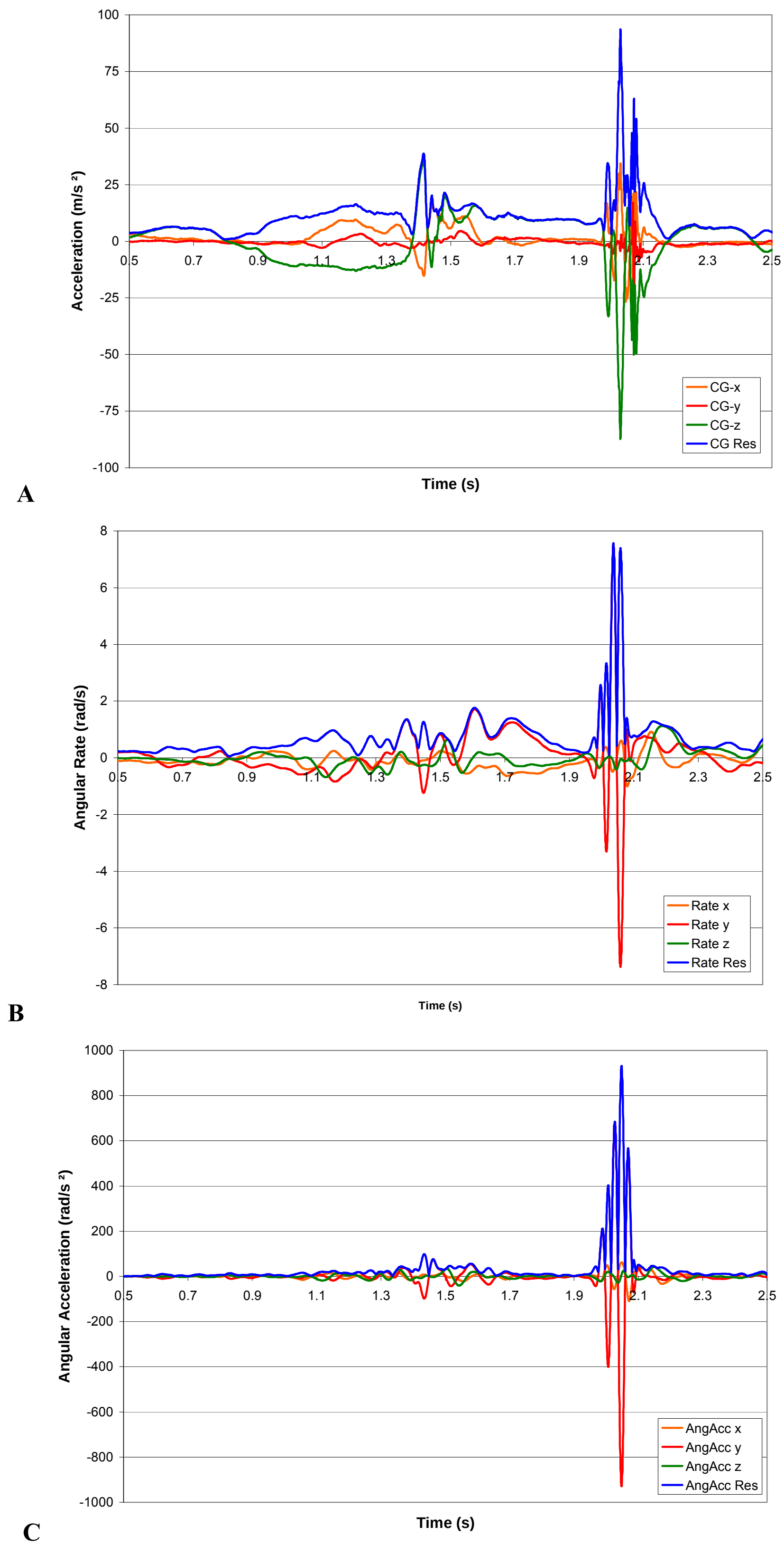

Figure 10. Linear acceleration, angular rate, and angular acceleration traces for maximum event from present study.

Center of gravity linear acceleration traces were analyzed for 3ms peak values. Clipped maximum positives and maximum negatives were almost identical to the unclipped results, with most deviating by less than 0.1 m/s^2. Because the values were so close, clipped values are not separately presented here.

Overall average peak angular rate data is shown below in Table 5 and Figure 11. As positive and negative for angular rate is simply a matter of polarity with little clinical relevance, the largest value was selected for graphing purposes. Angular rates were low as a rule, with no single direction greater than 5 rad/s and no resultant greater than 6 rad/s. Rates only surpassed 2 rad/s in three activities – Head Bob, Look Left, and Maximum Vertical Leap. ω_x and ω_y were usually greater than ω_z. Much like Y-axis acceleration, ω_z was much higher in Look Left than in the other tests. Maximum Vertical Leap had the largest angular rate response of the non-rotational events with ω_y above 2 rad/s and a resultant above 3 rad/s.

Once again, the overall maximum angular rates are appreciably larger than the average peaks – often two to three times as large. General trends again hold, although Head Nod replaces Look Left as the event with the largest angular rate response. Maximum Vertical Leap is once again the largest non-angular event response. An interesting side-effect of taking maximums instead of averages is that the resultant rotational rate is now much closer to the largest component rate than it was with the average of peaks. The maximums are shown in Table 6 and Figure 12.

Table 5. Overall average peak angular rates per event (rad/s)

Event		ω x	ω y	ω z	ω Res
1	Max	0.22	0.65	0.32	0.91
	Min	-0.29	-0.78	-0.31	0.02
2	Max	0.20	0.60	0.27	0.76
	Min	-0.34	-0.62	-0.27	0.02
3	Max	0.32	1.19	0.45	1.71
	Min	-0.56	-1.43	-0.43	0.03
4	Max	0.27	3.75	0.30	4.48
	Min	-0.42	-4.25	-0.40	0.05
5	Max	1.10	0.53	5.29	5.85
	Min	-1.22	-0.61	-5.18	0.05
6	Max	0.30	0.52	0.34	0.68
	Min	-0.48	-0.37	-0.42	0.02
7	Max	0.59	1.25	0.62	1.42
	Min	-0.76	-0.98	-0.67	0.04
8	Max	0.55	1.43	0.60	1.62
	Min	-0.70	-1.26	-0.70	0.04
9	Max	0.55	1.46	0.61	1.65
	Min	-0.70	-1.30	-0.71	0.04
10	Max	0.56	2.64	0.59	3.19
	Min	-0.75	-2.48	-0.70	0.03
11	Max	0.40	1.48	0.33	1.85
	Min	-0.55	-1.48	-0.40	0.03
12	Max	0.87	1.15	1.19	1.92
	Min	-1.10	-0.99	-1.10	0.04
13	Max	0.53	1.64	0.69	1.81
	Min	-0.83	-0.96	-0.73	0.04

Table 6. Overall maximum angular rates per event (rad/s)

Event		ω x	ω y	ω z	ω Res
1	Max	0.75	1.27	0.98	1.50
	Min	-0.79	-1.46	-1.07	0.00
2	Max	0.42	1.52	0.72	1.53
	Min	-0.69	-1.45	-0.52	0.00
3	Max	1.13	3.06	2.01	3.42
	Min	-2.07	-3.35	-1.27	0.00
4	Max	0.76	7.83	0.90	9.02
	Min	-0.92	-9.00	-1.60	0.00
5	Max	2.48	1.54	8.75	8.80
	Min	-2.90	-2.47	-8.10	0.02
6	Max	0.42	0.77	0.56	1.01
	Min	-0.72	-0.61	-0.94	0.00
7	Max	1.13	3.13	1.27	3.14
	Min	-1.26	-2.05	-1.05	0.00
8	Max	1.18	3.41	2.57	3.47
	Min	-1.16	-2.96	-1.46	0.01
9	Max	1.18	3.41	2.57	3.47
	Min	-1.16	-2.96	-1.46	0.01
10	Max	1.55	7.56	1.12	7.57
	Min	-1.65	-7.38	-1.68	0.00
11	Max	1.74	2.92	1.01	4.01
	Min	-1.15	-3.96	-1.07	0.00
12	Max	2.10	3.32	3.16	3.71
	Min	-2.25	-2.25	-2.17	0.00
13	Max	0.99	3.29	1.73	3.36
	Min	-1.78	-1.85	-2.26	0.01

Angular acceleration data is shown below in Table 7 and Figure 13. As positive and negative for angular acceleration is simply a matter of polarity, the largest value was selected for graphing purposes. Angular accelerations were low relative to injury criteria, with no single component or resultant greater than 250 rad/s^2. Angular accelerations only surpassed 150 rad/s^2 in two activities – Maximum Vertical Leap and Stair Jump. α_y tends to dominate the accelerations, as the highest single component in all activities except for Look Left, where α_z dominates. Maximum Vertical Leap again had the largest response, with a α_y of 222.2 rad/s^2 and a resultant of 230.4 rad/s^2.

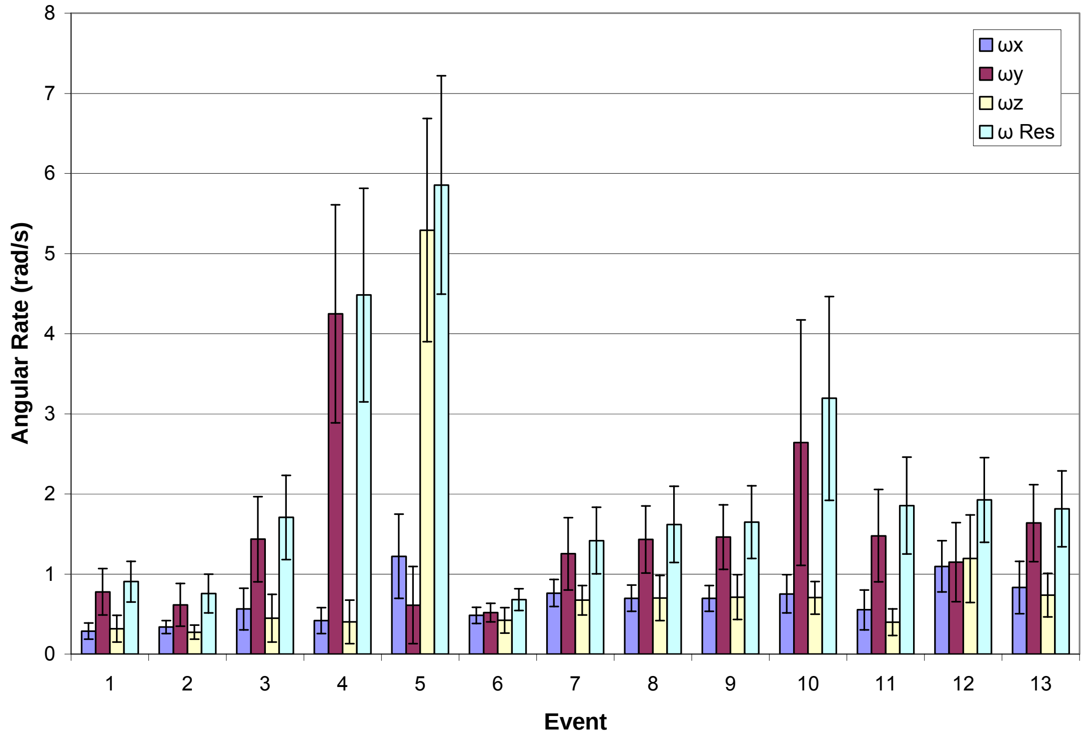

Figure 11. Peak angular rates per event. Results are mean ± standard deviation.

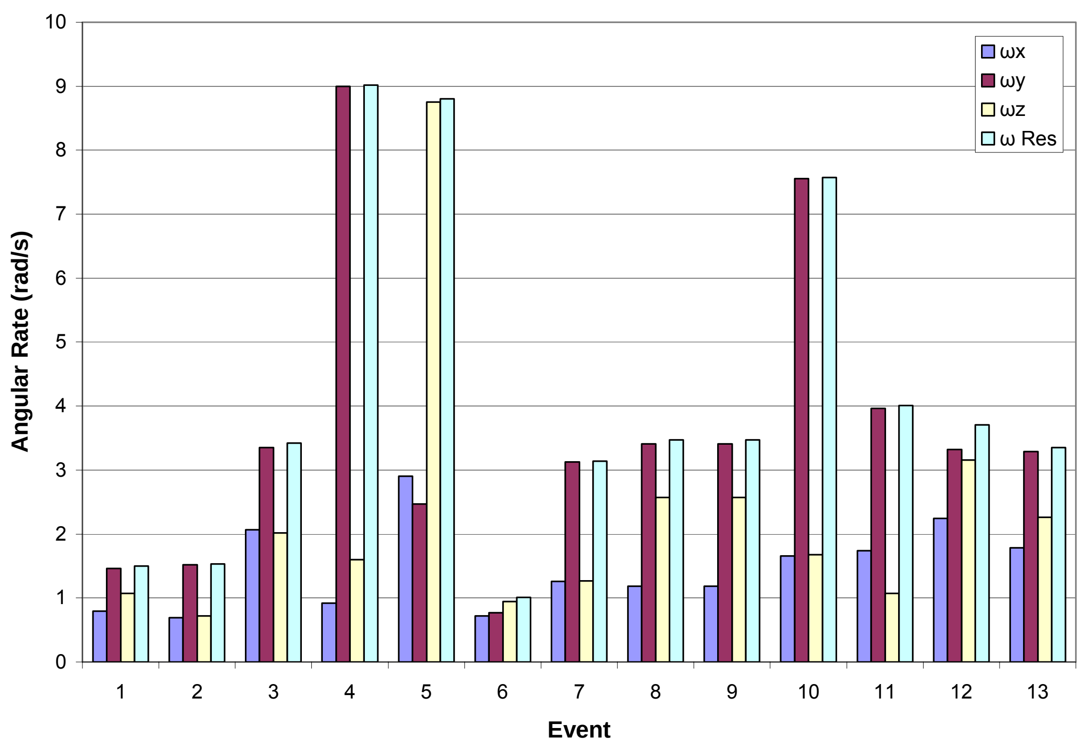

Figure 12. Maximum angular rates per event.

Table 7. Overall average peak angular accelerations per event (rad/s²)

Event		α x	α y	α z	α Res
1	Ave +	6.76	16.76	6.06	19.26
	Ave -	-6.66	-15.08	-6.00	
2	Ave +	6.02	10.83	4.60	13.31
	Ave -	-5.82	-11.91	-4.59	
3	Ave +	21.07	64.53	11.57	84.52
	Ave -	-21.72	-77.12	-13.06	
4	Ave +	7.98	37.56	6.69	73.78
	Ave -	-7.94	-72.81	-8.12	
5	Ave +	17.32	15.84	54.14	87.59
	Ave -	-19.04	-14.20	-85.87	
6	Ave +	11.57	16.05	9.45	24.88
	Ave -	-12.84	-20.96	-7.16	
7	Ave +	24.74	50.57	14.46	66.06
	Ave -	-25.72	-62.37	-15.17	
8	Ave +	29.94	74.34	20.12	85.63
	Ave -	-28.39	-73.74	-19.91	
9	Ave +	30.24	74.93	20.89	85.78
	Ave -	-28.81	-73.91	-20.22	
10	Ave +	47.39	173.31	25.75	230.36
	Ave -	-49.92	-222.17	-27.23	
11	Ave +	36.91	112.42	13.33	151.00
	Ave -	-32.84	-142.74	-14.43	
12	Ave +	46.38	46.15	29.83	78.34
	Ave -	-47.33	-54.86	-26.87	
13	Ave +	29.05	59.16	19.78	79.92
	Ave -	-29.02	-69.72	-18.49	

Resultant angular accelerations cannot be negative.

Table 8. Overall maximum angular accelerations per event (rad/s²)

Event		α x	α y	α z	α Res
1	Max +	27.74	41.62	16.13	43.56
	Max -	-24.84	-38.29	-13.04	
2	Max +	12.27	32.10	14.21	52.58
	Max -	-18.35	-51.62	-9.72	
3	Max +	72.49	229.00	58.67	320.25
	Max -	-79.66	-308.59	-53.37	
4	Max +	26.12	98.37	15.83	183.09
	Max -	-24.29	-181.50	-43.94	
5	Max +	51.96	64.84	107.90	216.19
	Max -	-55.68	-40.50	-213.92	
6	Max +	19.88	30.98	29.83	48.62
	Max -	-22.82	-39.42	-10.87	
7	Max +	38.93	140.01	26.50	208.70
	Max -	-48.45	-208.10	-23.14	
8	Max +	77.92	225.74	85.31	232.58
	Max -	-54.77	-232.39	-70.22	
9	Max +	77.92	225.74	85.31	232.58
	Max -	-54.77	-232.39	-70.22	
10	Max +	147.54	682.98	55.48	931.25
	Max -	-182.36	-929.15	-73.08	
11	Max +	184.54	346.01	30.05	445.09
	Max -	-149.86	-439.71	-53.37	
12	Max +	109.68	99.41	81.07	173.50
	Max -	-140.43	-133.87	-89.99	
13	Max +	77.05	184.99	48.21	239.81
	Max -	-73.31	-237.85	-40.26	

Resultant angular accelerations cannot be negative.

Maximum angular accelerations are shown in Table 8 and Figure 14. Maximum Vertical Leap shows the largest response again, with a α_y and resultant of almost 1000 rad/s^2. Plop Sit shows a larger maximum than would be expected from its averaged results, with an α_y over 300 rad/s^2, making it the third largest event, after the two jumping activities. Maximums are again approximately three times larger than their averaged counterparts.

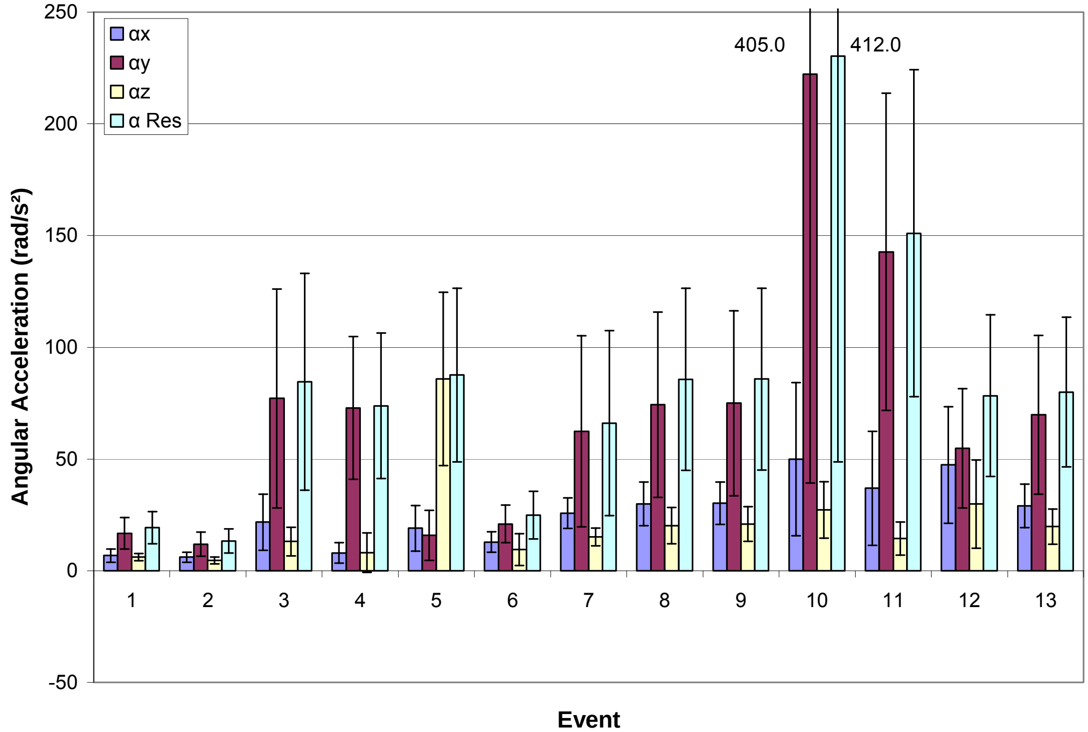

Figure 13. Average peak angular accelerations per event. Results are mean ± standard deviation.

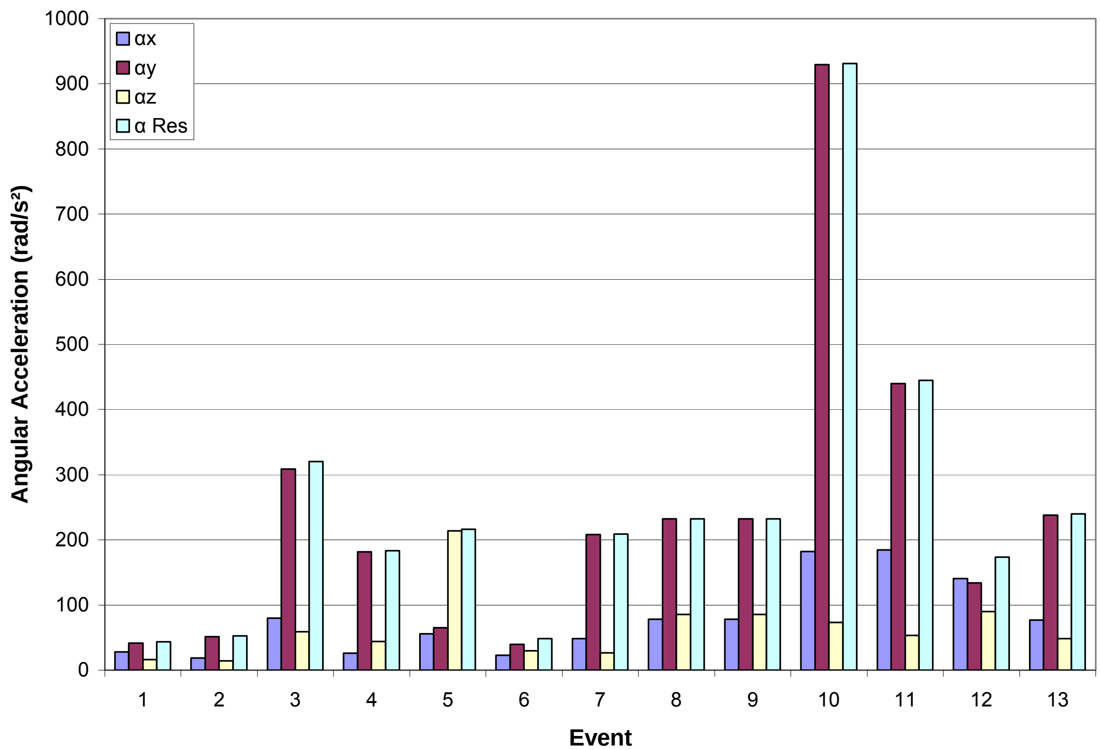

Figure 14. Maximum angular accelerations per event.

2.3.2 Gender Differences

The maximum responses by gender have similar trends to averages by gender. Males tend to have higher accelerations and rates in the jumping and throwing activities, as well as Look Left, whereas females tend to have higher accelerations and rates in the sitting tasks, walking/running, jumping jacks. These results are shown in Tables 9 and 10, and Figures 15-17.

Data between the sub-groups was also compared for statistically significant differences, although these results are not reported due to the limited number of samples (4 or 5) within each group. A lack of gender- or size-specific trends and the occurrence of statistical oddities within the comparisons reinforce this decision. For instance, the small female and large male groups tended to not have significant differences, whereas the mid-male and mid-female groups tended to have statistically significant differences.

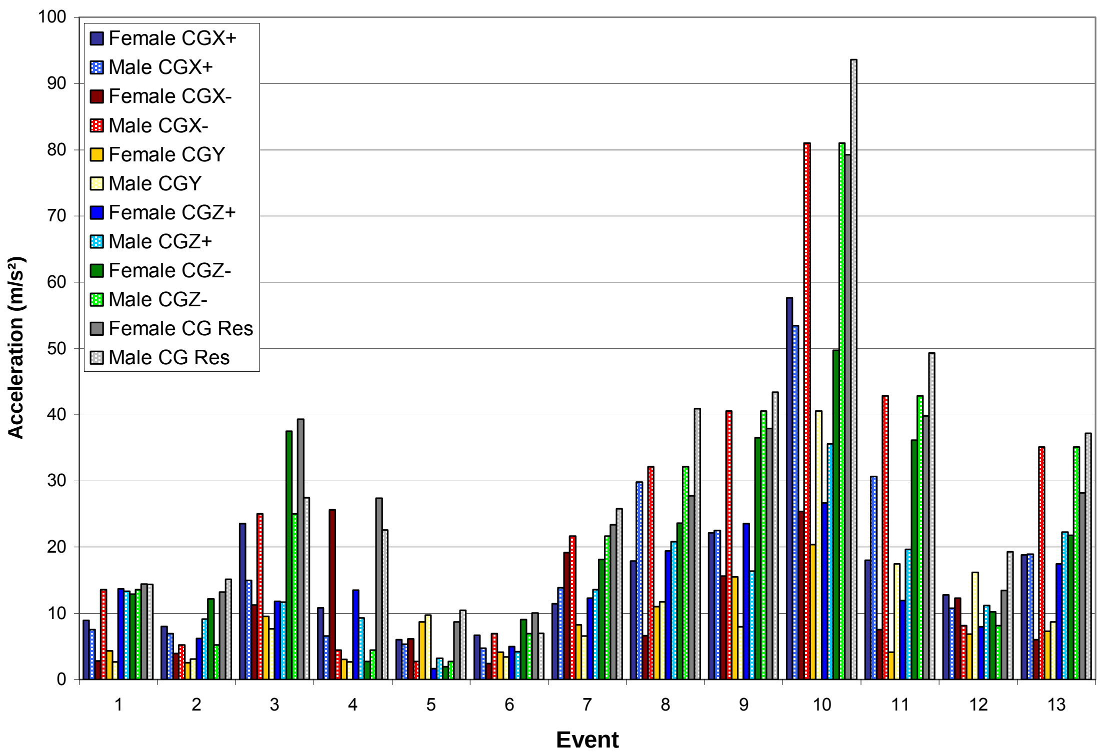

Figure 15. Male vs female maximum linear accelerations at the cg.

Table 9. Male maximum responses.

Males	1	2	3	4	5	6	7	8	9	10	11	12	13
ω x	0.40	0.48	0.85	0.89	2.90	0.72	1.26	1.17	1.17	1.65	1.74	2.10	1.78
ω y	1.33	1.46	2.44	7.95	2.47	0.77	2.12	2.68	2.68	7.56	3.96	3.32	3.29
ω z	0.85	0.52	0.68	1.60	8.75	0.57	1.27	1.22	1.22	1.68	1.07	3.16	2.26
ω Res	1.33	1.48	2.45	8.00	8.80	0.87	2.13	2.69	2.69	7.57	4.01	3.71	3.33
α x	18.21	9.51	59.84	26.12	55.68	13.87	40.08	62.76	62.76	182.36	184.54	140.43	45.28
α y	38.29	19.73	195.57	181.50	64.84	32.46	161.56	187.91	187.91	929.15	439.71	125.50	237.85
α z	16.13	11.03	24.87	43.94	213.92	13.58	23.27	67.92	67.92	73.08	53.37	89.99	48.21
α Res	39.76	19.74	197.00	183.09	216.19	32.52	162.32	189.11	189.11	931.25	445.09	173.50	239.81
CGX+	7.55	6.94	14.97	6.57	5.35	4.77	13.87	29.83	22.53	53.44	30.70	10.77	18.92
CGX-	7.97	12.95	15.11	21.56	8.32	3.51	13.11	7.69	15.29	52.84	16.78	18.22	8.95
CGY	2.65	3.10	7.70	2.66	9.77	3.41	6.60	11.76	7.98	40.51	17.50	16.21	8.68
CGZ+	13.31	9.16	11.67	9.34	3.23	4.18	13.58	20.79	16.39	35.63	19.65	11.18	22.26
CGZ-	13.62	5.32	25.21	4.50	2.85	7.01	22.03	32.69	40.56	87.21	47.02	8.36	36.71
CG Res	14.39	15.13	27.42	22.58	10.48	7.02	25.82	40.90	43.38	93.62	49.28	19.30	37.22

Table 10. Female maximum responses.

Females	1	2	3	4	5	6	7	8	9	10	11	12	13
ω x	0.79	0.69	2.07	0.92	2.25	0.64	1.23	1.18	1.18	1.43	0.92	2.25	1.41
ω y	1.46	1.52	3.35	9.00	1.03	0.63	3.13	3.41	3.41	5.89	2.70	2.25	3.10
ω z	1.07	0.72	2.01	0.85	7.44	0.94	1.00	2.57	2.57	1.20	1.01	2.27	1.76
ω Res	1.50	1.53	3.42	9.02	7.78	1.01	3.14	3.47	3.47	5.90	2.75	2.57	3.36
α x	27.74	18.35	79.66	24.29	37.82	22.82	48.45	77.92	77.92	141.45	64.35	87.96	77.05
α y	41.62	51.62	308.59	120.60	27.53	39.42	208.10	232.39	232.39	661.85	257.30	133.87	133.29
α z	13.99	14.21	58.67	13.53	129.32	29.83	26.50	85.31	85.31	64.46	25.03	43.04	38.21
α Res	43.56	52.58	320.25	121.10	130.84	48.62	208.70	232.58	232.58	663.60	260.27	136.24	135.31
CGX+	8.94	8.05	23.53	10.83	6.04	6.69	11.46	17.87	22.16	57.62	18.02	12.81	18.83
CGX-	2.79	3.95	11.28	25.65	6.18	2.41	19.17	6.65	15.66	25.37	7.55	12.29	6.03
CGY	4.30	2.56	9.55	3.06	8.70	4.14	8.29	11.01	15.50	20.42	4.16	6.88	7.31
CGZ+	13.71	6.20	11.78	13.54	1.67	4.99	12.32	19.44	23.57	26.67	11.90	8.00	17.44
CGZ-	13.15	12.22	37.79	2.93	1.97	9.17	18.19	23.85	36.73	60.67	37.57	10.29	22.05
CG Res	14.41	13.18	39.35	27.41	8.71	10.05	23.37	27.74	37.92	79.27	39.78	13.44	28.21

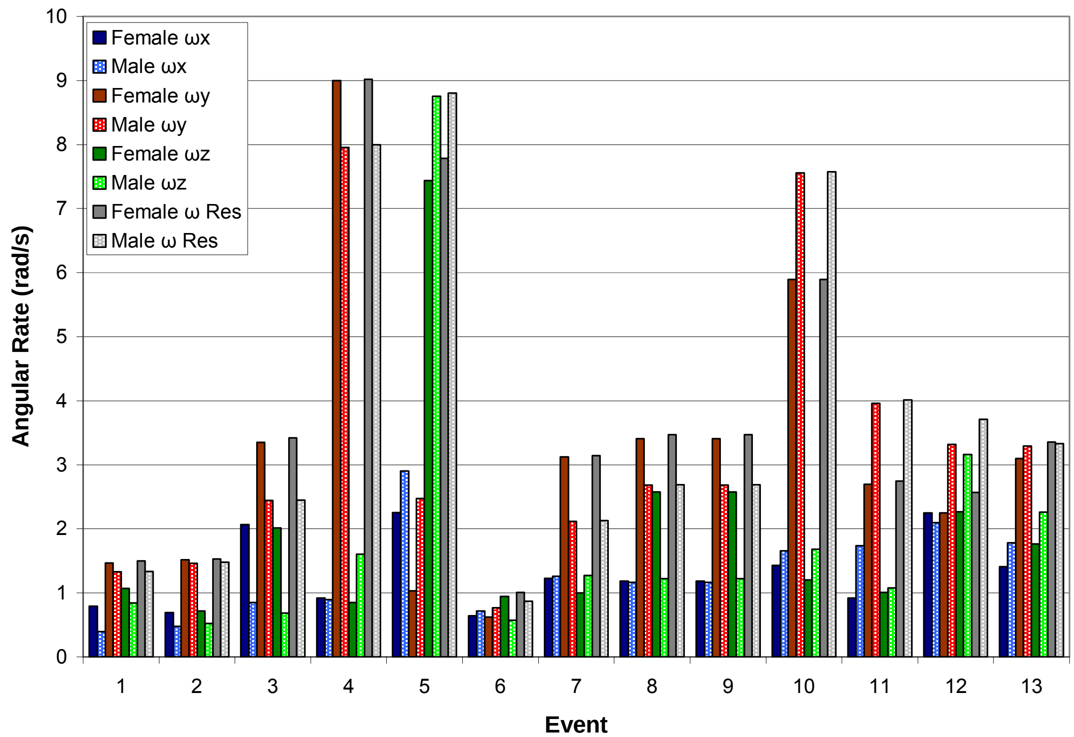

Figure 16. Male vs female maximum angular rates.

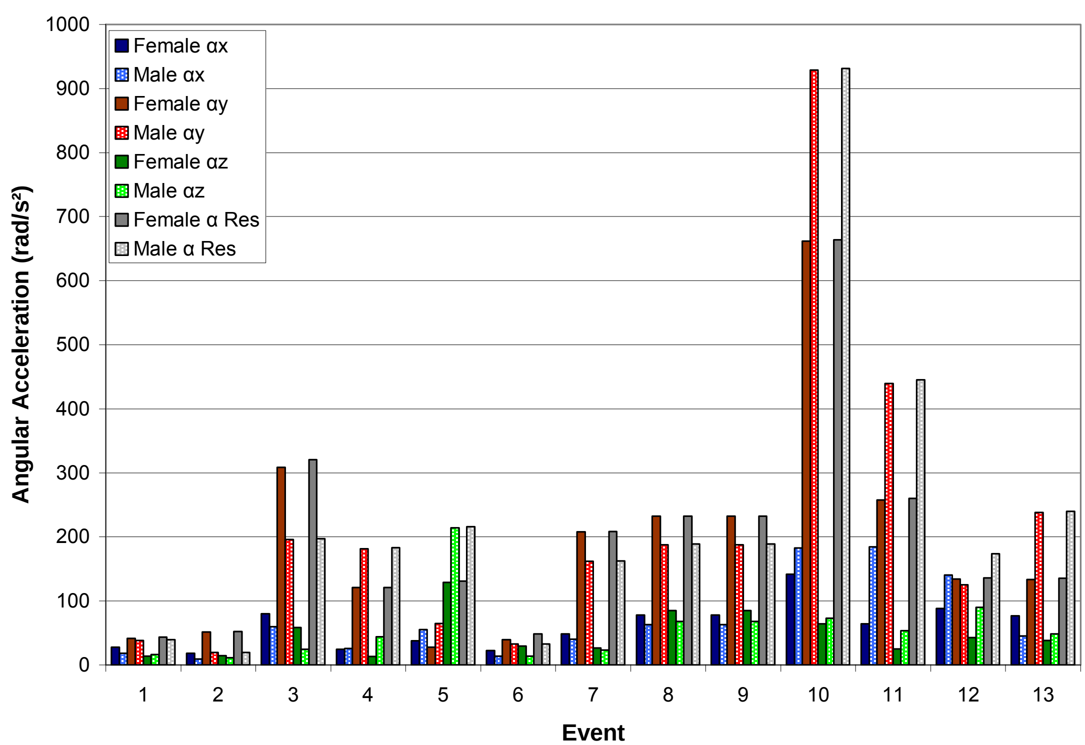

Figure 17. Male vs female maximum angular accelerations.

2.3.3 Statistical Differences

There were significant or partially significant differences in average peak responses between the male and female volunteers, with significance defined as a matched Student's T-Test result less than 0.05 and partial significance as a result less than 0.10, in 21.9% (60/273) of comparisons. Only Run in Place and Jumping Jacks exhibited no statistically significant differences between genders. Males tended to have larger linear accelerations, especially in jumping or shooting/throwing activities, although females tended to have the higher accelerations in sitting and walking/running. Differences were often significant in these events, although no acceleration difference was significant for Run, Run in Place, or Jumping Jacks. Angular rate and acceleration results are much the same as linear acceleration. Males tend to have higher responses in jumping and throwing/shooting activities, whereas females tend to have higher responses in walking/running and sitting. Males had significantly higher responses for the rotational events of Look Left, Walk, Run, Maximum Vertical Leap, Stair Jump and Throw Ball, especially rotating around the z-axis. A breakdown of significance by event is shown in Table 11, with comparisons of acceleration, angular rate and angular acceleration in Figures 18-20.

Table 11. Statistical differences between average peak results in male and female volunteers based on matched T-Test.

Event	**1**		**2**		**3**		**4**		**5**	
Significance	*Max +*	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*
ω x	0.159	**0.036**	0.953	0.309	0.130	*0.099*	0.892	0.483	**0.007**	0.119
ω y	0.673	0.482	0.203	0.706	0.271	0.765	0.965	*0.072*	0.638	0.176
ω z	0.412	0.143	0.969	0.342	0.180	0.114	0.412	0.225	0.330	**0.027**
ω Res	0.402		0.268		0.958		0.654		*0.056*	
α X	0.146	0.143	0.594	0.955	0.118	*0.095*	0.842	0.679	**0.011**	**0.024**
α Y	0.798	*0.063*	0.327	0.374	0.383	0.922	0.208	0.348	*0.066*	0.166
α Z	0.464	0.537	0.632	0.666	*0.053*	0.426	0.345	**0.033**	0.122	**0.017**
α Res	0.131		0.250		0.471		0.107		*0.072*	
CG X	0.192	**0.000**	0.555	**0.014**	0.111	**0.001**	0.208	0.409	0.182	**0.005**
CG Y	0.689	*0.073*	0.296	0.187	0.402	0.554	0.475	0.518	0.113	**0.020**
CG Z	0.867	0.580	0.754	*0.087*	0.517	0.149	0.888	*0.052*	0.176	**0.039**
CG Res	0.760		0.624		0.169		0.547		0.192	
Event	**6**		**7**		**8**		**9**			
	Max +	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*		
ω x	0.221	0.442	**0.005**	**0.034**	0.273	0.277	0.310	0.263		
ω y	0.983	0.793	0.829	0.199	0.270	0.501	0.410	0.742		
ω z	0.996	0.303	0.851	0.744	0.433	0.422	0.571	0.521		
ω Res	0.597		0.361		0.319		0.433			
α x	**0.035**	**0.028**	*0.053*	**0.038**	0.115	0.324	0.155	0.425		
α y	0.105	0.231	0.296	0.506	0.368	0.453	0.398	0.464		
α z	0.286	0.253	0.118	0.363	0.723	0.716	0.953	0.825		
α Res	**0.032**		0.290		0.308		0.315			
CG X	0.592	**0.032**	0.586	0.380	0.561	0.403	0.508	0.282		
CG Y	0.919	**0.003**	0.288	0.430	0.446	0.602	0.325	0.686		
CG Z	0.294	0.122	0.168	0.679	0.491	0.617	0.625	0.941		
CG Res	0.079		0.558		0.623		0.850			
Event	**10**		**11**		**12**		**13**			
	Max +	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*	*Max +*	*Max -*		
ω x	0.179	0.370	0.808	0.297	0.301	0.166	0.735	0.963		
ω y	0.192	0.528	**0.026**	0.417	0.847	0.316	*0.069*	0.300		
ω z	*0.083*	**0.024**	0.220	0.411	0.585	*0.071*	0.653	0.457		
ω Res	0.369		**0.042**		0.260		0.649			
α x	0.437	0.516	0.438	0.461	**0.031**	0.107	0.593	0.760		
α y	0.328	0.297	0.165	**0.022**	0.067	0.174	*0.083*	*0.098*		
α z	**0.031**	0.295	0.187	0.209	**0.027**	**0.008**	0.446	0.779		
α Res	0.266		0.103		**0.026**		*0.078*			
CG X	0.191	**0.008**	0.162	**0.000**	0.829	**0.001**	0.678	**0.025**		
CG Y	0.146	0.114	**0.034**	**0.014**	**0.027**	**0.026**	0.790	0.105		
CG Z	0.211	**0.041**	0.370	0.129	0.328	0.863	0.634	**0.030**		
CG Res	*0.054*		*0.090*		*0.053*		0.117			

Resultants cannot be negative. Values significant to $T < 0.05$ are **bolded**. Values significant to $T < 0.10$ are *underlined italic*.

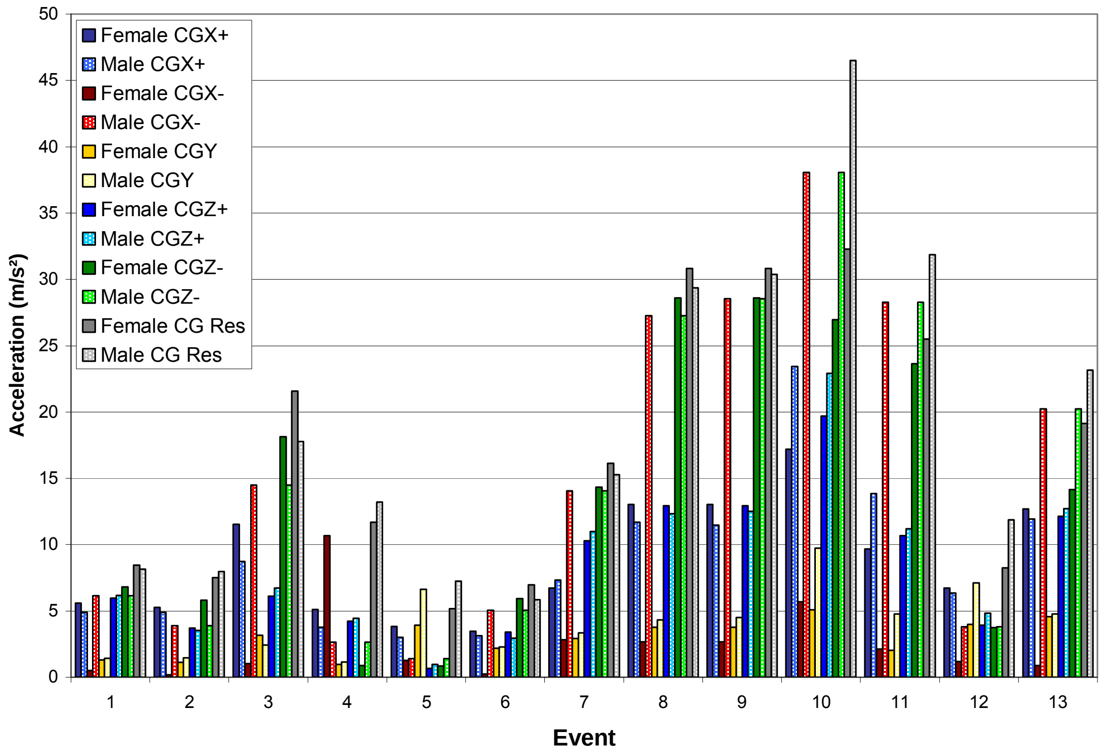

Figure 18. Male vs female average peak linear accelerations at the cg.

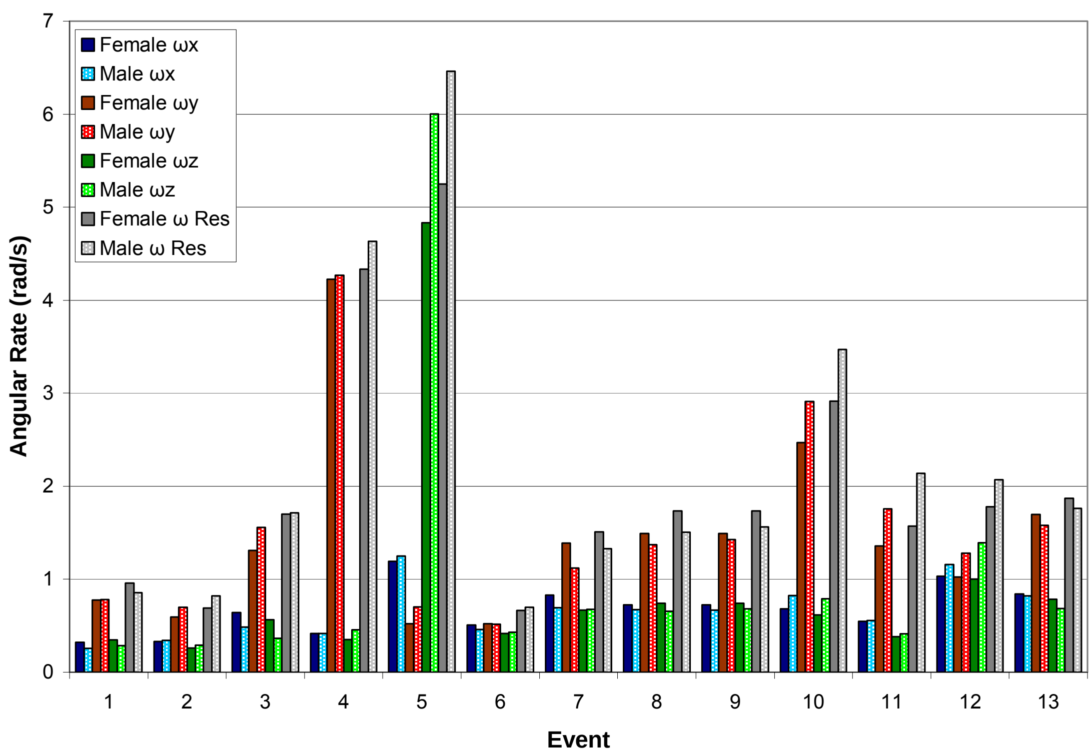

Figure 19. Male vs female average peak angular rates.

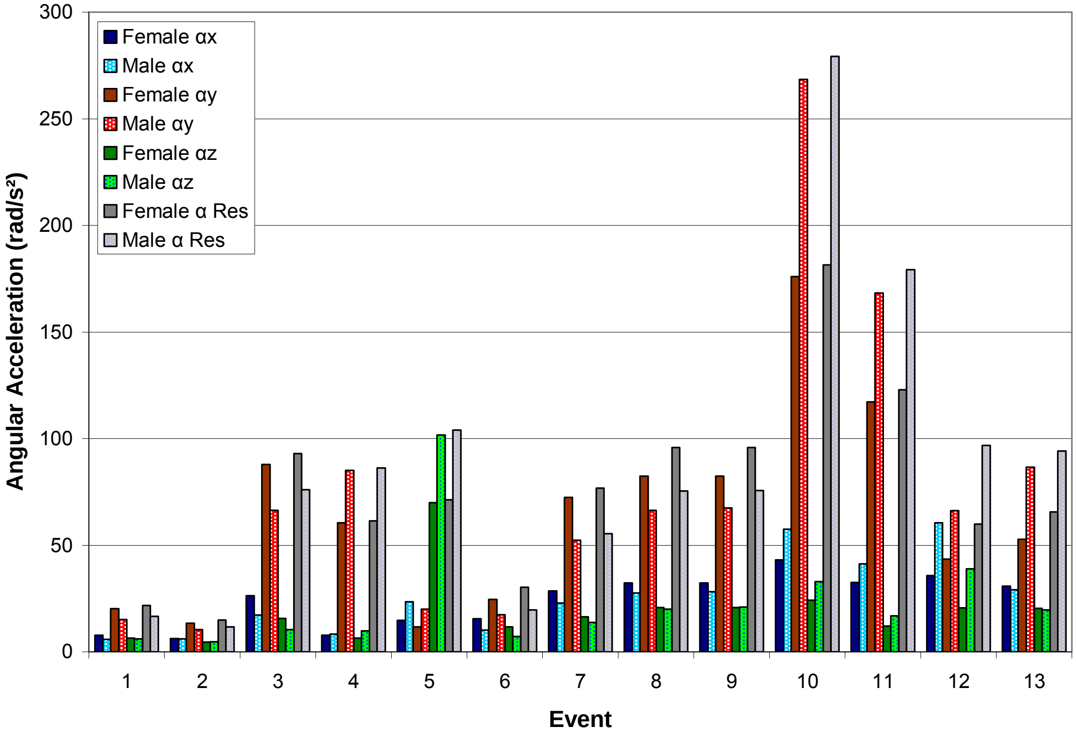

Figure 20. Male vs female average peak angular accelerations.

2.4 DISCUSSION

2.4.1 Average and Maximum Responses

The accelerations reported are all of a fairly low level. No subjects complained of discomfort or described the activities as especially rigorous. The highest resultants over the course of 700 recordings were only approximately 9.5 G, 9 rad/s and 900 rad/s^2. This only results in an unbounded HIC of 3.1, with HIC_{36} equal to 2.5 and HIC_{15} equal to 2.3. The maximum resultants from this test series are only one-third the average linear acceleration and approximately equal to the ω_x seen in an average, non-injurious football collision (Naunheim et al 2003, Duma et al 2005). Results are significantly lower than the threshold for a 75% MTBI risk established by King – 333 HIC, 98 G and 7130 rad/s^2 (King et al 2003). These maximums are similar to the 7g mean peak head accelerations seen in 14.5 km/hr whiplash recreations, however (Allen et al 1994). Pulse durations were even similar, as the acceleration pulses of this study were typically in the 100-200ms range. It is difficult to compare to whiplash, however, as while head accelerations can be similar, neck kinematics are often far different. Also, this series of tests tended to have higher Z-axis accelerations than X-axis accelerations, unlike whiplash scenarios.

Common events compare well with the previous literature. Stand Up, Sit Down, and Head Nod agree with the corresponding events in Allen, despite Head Nod and Head Bob occurring in opposite directions. Allen's Plop in Chair and Jump off Stair had a more forceful response than this study, but this study had the more forceful Look Left and Head Nod. It should be noted that subjects in this study were instructed to execute the Look Left and Head Nod actions more quickly than Allen's were, and this likely

explains the difference. Largest differences were between this study's and Allen's Stair Jump and Chair Plop. This difference is likely attributable to differences in technique or effort as opposed to differences in instruction or material differences in the step or chair. The addition of the foam pads on the subject's spines may have impacted the x-axis accelerations for Chair Plop as well, as subjects often avoided contacting the chair back when sitting due to discomfort from hitting the pads and accelerometers. Differences in accelerometer placement between the Allen study and this study, as explored in the second part of this thesis, are not sufficient to explain the acceleration differences in these events. The common events are shown in Figure 21.

Walking and running are daily activities that have been previously reported as well. Kavanagh and Woodman reported the average linear accelerations in walking, and Woodman also reported average angular accelerations (Kavanagh et al 1996, Woodman and Griffin 2004). Mercer and Mahar both reported linear accelerations in running. Results from this study agreed very well with the previous literature for both walking and running, achieving almost identical results for linear acceleration (Mahar et al 1997, Mercer et al 2003). Angular accelerations for walking were approximately 50% larger in this study than in Woodman. While it is possible that the weight of the mouthpiece induced the extra angular acceleration seen in this study, it should be noted that Woodman's subjects wore a 1.3 kg helmet as well as an instrumented bite bar. It is more likely attributable to differences of calculation method. Woodman calculated angular accelerations using differences in two accelerometers whereas this study differentiated angular rates (Woodman and Griffin 2004). Linear acceleration comparisons are in Figure 22 and angular accelerations are in Figure 23.

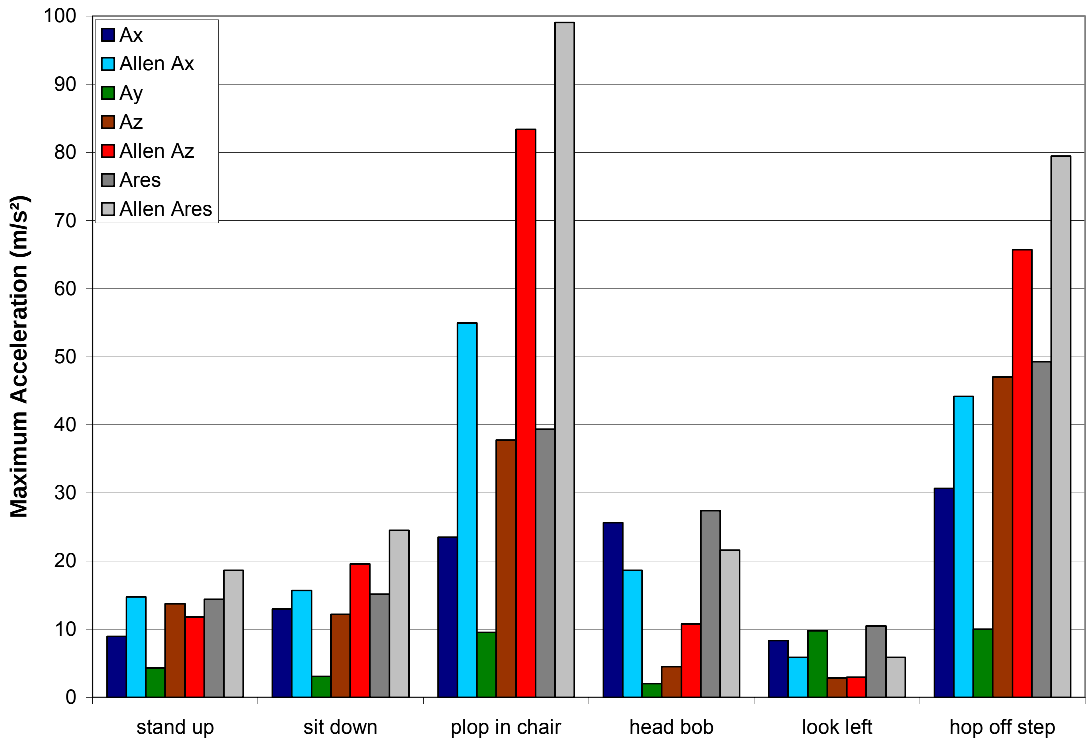

Figure 21. Comparison of common events from Allen (1994) and this study.

It is somewhat harder to compare this study's results with those of Arndt in the Exponent study (Exponent 2002, Arndt et al 2004), as there are no common activities. Some activities were comparable, however. Run in Place and Jumping Jacks were similar to the obstacle course and pogo stick activities. Maximum Vertical Leap was similar in magnitude to falling down and about half that of the pillow strike event. It is interesting to note that Plop in Chair, Run in Place, Stair Jump, and Jumping Jacks all had maximum linear accelerations and pulse durations similar to that of the rollercoasters in prior studies (Exponent 2002, Smith and Meaney 2004). The lack of a one-to-one correspondence makes it more difficult to extrapolate to the potential angular rates and accelerations in these events, however.

With the general agreement between linear acceleration data from this study and previous daily activity literature, as well as the agreement between walking angular accelerations with Woodman, it is reasonable to conclude that the data from this study is consistent with the general trends of these activities. It is also reasonable to extrapolate the data from this study to fill in the gaps in the previous literature, specifically that this study's angular rates and accelerations are also likely in agreement with the unmeasured rates and accelerations in the literature. This study reports for the first time the angular rates and angular accelerations of the head for everyday events in a test series consistent in result with previous literature.

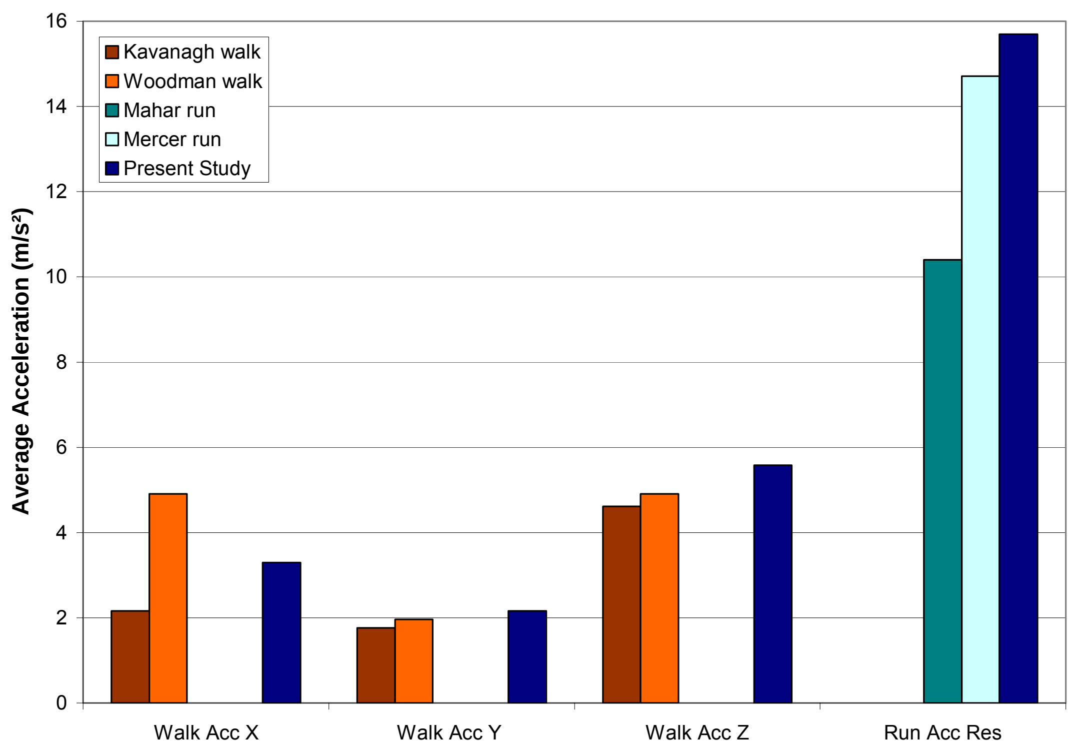

Figure 22. Comparison of linear accelerations in walking and running between this study and previous literature.

It is difficult to estimate how well the results of this study could be applied to a juvenile or elderly population. It is likely that either group would have generated less forceful responses than the young adult population in this study, but it is not clear whether accelerations and rates from this data set would prove injurious to either population. It is likely that an elderly population would have a reduced acceleration tolerance, but it is harder to estimate the tolerance of a juvenile population. This is a subject which should be investigated in the future.

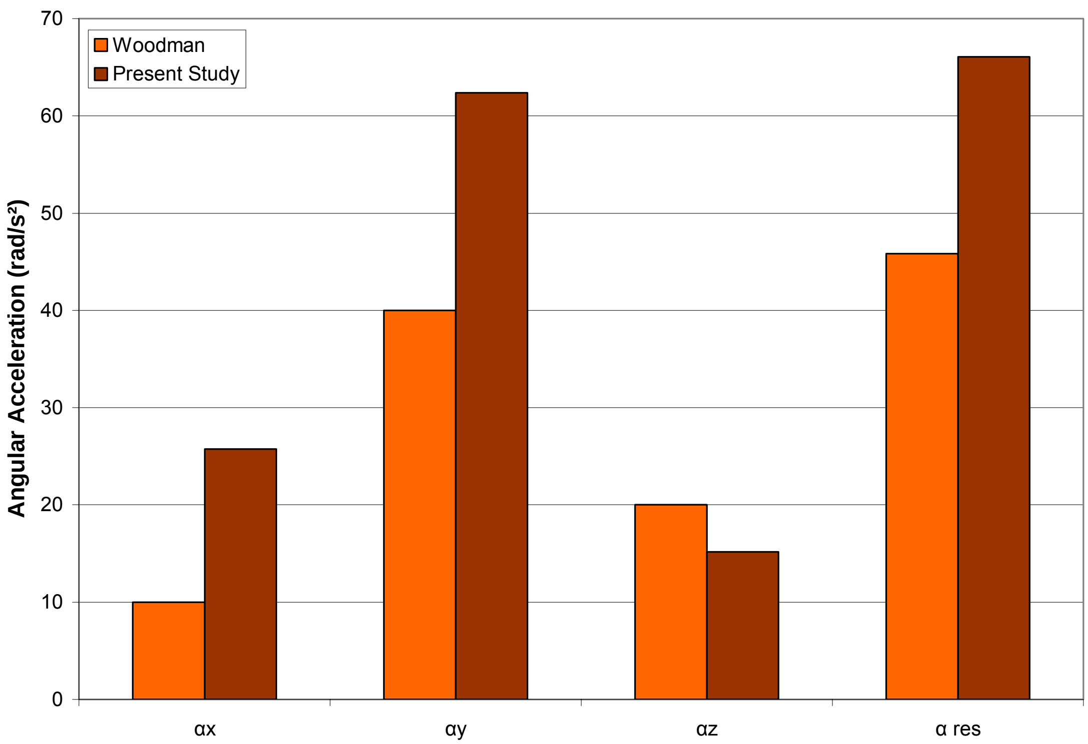

Figure 23. Comparison of angular accelerations in walking between this study and Woodman and Griffin 1996.

2.4.2 Statistical Differences

It is clear that there were many statistically significant differences between male and female responses in this study. What is less clear is whether these differences are of any clinical importance or even whether they are actually related to gender at all.

Statistical differences tended to occur more often in the lower level activities, with maximum accelerations less than 4 G. As injury at this level is very unlikely, it is an open question whether there is any practical difference between a male response of 4 G and a female response of 3 G. It appears that the primary reason the more forceful activities showed fewer gender-related statistical differences is their considerable variation from subject to subject. Statistical differences were lost in the large individual variations.

Individual variation may be the most important parameter. While subjective and anecdotal, there was a pronounced difference in effort involved in some activities. One explanation for the tendency of male responses to be higher is that most of the subjects perceived as trying the hardest were male. Most likely, the reason male responses were higher for Maximum Vertical Leap is that the male subjects, on average, jumped higher. While some of this is athletically based, there were notable differences in the effort of subjects in this activity, and males, on average, tried harder in this activity. This study is not equipped to determine whether this indicates an underlying psychological factor which could show a gender factor. Regardless, an effort factor would explain why male responses tended to be higher in self-regulating activities like Maximum Vertical Leap, whereas female responses tended to be slightly higher in normalized activities, such as sitting and treadmill walking and running. Stride length differences could be a factor in the treadmill activities, however, with female subjects having to work harder, relative to their larger male counterparts, to maintain the same speed. In theory, mass differences between male and female subjects could also play a role in gender differences, but this too seems to be lost in the effort effect. An interesting result was that there were fewer

statistically significant differences between the small female and large male categories, with an average difference of 9.3 in and 86.4 lb, than there were between the mid-male and mid-female categories, with an average difference of 3.5 in and 27.8 lb. More likely is the explanation that with just four or five data points per category, that individual effort variation played a more significant role than size or gender variation.

As mentioned previously, this individual effort difference effect can also explain some of the disparities when compared to Allen's data, where some tasks were approximately equal, some were higher in this study and some were pronouncedly higher in Allen. Much of this is most likely explain by differences in volunteer effort on those tasks between Allen and the present study.

2.4.3 Limitations

This study is limited by the age of the volunteers. As all volunteers were between the ages of 19 and 32, it is unknown how well this data correlates with safety thresholds in a juvenile or elderly population.

Despite a larger sample size than previous studies of this kind, this research is still limited by a relatively small sample. This limitation is especially pronounced when investigating differences between genders and subgroups. It is desirable for future testing to include more subjects and subjects from a wider age range, preferably including pre-teen and elderly subjects.

A consequence of volunteer testing in dynamic, self-generated activities is the inability to rigorously control volunteer effort. Although this research attempted to regulate subject effort level, there was no objective measure involved. Although perhaps

an insurmountable problem, it is desirable for future testing to include a method to control the level of volunteer effort.

Any in vivo measurement of head acceleration in human subjects has limitations. Helmet systems are limited by helmet slip, large offsets from the center of gravity, and noise from the helmet response. Accelerometers attached to the periphery of the head need to overcome relative motion problems. Accelerometer arrays mounted to the head have to overcome skin slip and other problems. Biteplates, though recommended by King in a comment on the Allen paper (King 1994), are limited by the ability of the jaw to maintain the plate in a constant position. Some high frequency responses and some of the larger short-duration angular accelerations in this study may be a result of transient slip due to jaw movement. Also, the effect of the biteplate mass on head inertia is unknown as subject head inertias were not recorded due to measurement difficulty in vivo. It would be advantageous for future studies to account for these effects.

It is unknown whether subject bite strength and jaw load has an effect on results. The compliance of the mouthpiece system – impression tray, mouth guard and teeth – is unknown. It would be advantageous for future studies to directly measure these factors.

It is believed that the angular acceleration terms are not excessively noisy, despite being calculated from the derivative of the angular rates, in part because the results were filtered to a 30 Hz cut-off – a level chosen based on observation to preserve signal while eliminating noise. It would be advantageous for future studies to performed Fourier analysis of the resulting signals, however, to determine more analytically the amount of noise remaining and their frequencies.

2.5 CONCLUSION

A total of 18 young adult subjects – 9 male and 9 female – were instrumented via a biteplate system to record linear acceleration, angular rate, and angular acceleration while performing 3 repetitions of 13 everyday activities for a total of 700 records. Head center of gravity accelerations were derived from mouth accelerometer traces and angular acceleration data was derived by numerically differentiating angular rates.

Six degree-of-freedom head acceleration data from everyday activities was reported for the first time. Linear accelerations and angular rates agreed well with previous studies, which allow the results of this study to fill the neglected data from prior studies. Maximum Vertical Leap was the most forceful activity measured for linear and angular acceleration, with a resultant linear acceleration of 93.6 m/s^2, a resultant angular acceleration of 931.25 rad/s^2. Head Nod demonstrated the largest angular rate, at 9.02 rad/s. Maximum resultant linear accelerations were typically less than 50 m/s^2, maximum angular rates were typically less than 4 rad/s, and maximum angular accelerations were typically less than 300 rad/s^2 in the tested everyday activities.

Male and female data was often different to the level of statistical significance, although it could not be determined due to sample size and an inability to objectively regulate volunteer effort whether this was due to gender or size differences, or simply a matter of variation in individual effort.

2.6 REFERENCES

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Chapter 3

Effect of Angular Terms on Linear Acceleration in Daily Events

ABSTRACT

OBJECTIVE: The goal of this study was to determine the significance and magnitude of angular term influence on linear acceleration in daily activities.

STUDY DESIGN: Mouth array versus center of gravity accelerations were compared against the full 700 sample data set from the previous section. Two prior studies were compared using data from subject 11. Corrections were made to four comparable events from a prior study.

RESULTS: Average error was strongly influenced by the type of motion in each event, ranging from -3.1% to 115.2% when converting from mouthpiece resultant accelerations to center of gravity acceleration. Error increases as angular rates and accelerations increase. Mouth array accelerations are statistically significantly different than center of gravity accelerations. Across the same data set, prior studies' arrays were statistically significantly different equally often. Errors are somewhat less but still significant for the prior studies. Peak accelerations in Allen et al 1994 may have been off by as much as 2 G when converted to center of gravity accelerations.

CONCLUSIONS: This study showed that the effects of angular terms cannot be neglected when measuring head accelerations in daily activities.

3.1 INTRODUCTION

The medical and engineering literature contains many studies of single or multiple "everyday life" activities using volunteers (Allen et al 1994, Arndt et al 2004, Exponent 2002, Kavanagh et al 2004, Mahar et al1997, Mercer et al 2003, Smith and Meaney 2002, Woodman 1996). The majority of these studies concentrate on linear accelerations and either neglect or neglect to report the angular components – angular velocity and angular acceleration. The difficulty is that accelerations measured other than at the center of rotation, usually taken as the center of gravity, are influenced by these angular components. As all volunteer studies measure at non-central locations by necessity, all are potentially prone to this effect. Unfortunately, the combined effect of the angular components during low-level "everyday life" activities is not known.

The purpose of this study was to determine the importance of the angular components of acceleration in everyday activities, and as an example determine possible correction factors for previous literature.

3.2 METHODOLOGY

3.2.1 Data Comparison

From an earlier portion of this study, a total of 18 young adult subjects – 9 male and 9 female – were instrumented via a biteplate system to record linear acceleration, angular rate, and angular acceleration while performing 3 repetitions of 13 everyday activities for a total of 700 records. Head center of gravity accelerations were derived from mouth accelerometer traces and angular acceleration data was derived by numerically differentiating angular rates. From this data, it was possible to determine the difference between the linear accelerations at the mouth and at the center of gravity of the head. Peak accelerations in the three component directions, as well as peak resultant acceleration, were recorded for both the mouth and the resolved center of gravity accelerometers and compared for statistical significance. Statistical significance would indicate that the effects of angular rate and acceleration were significant for that event and cannot be neglected.

The accelerometer locations from two previous major studies of everyday activities were estimated (Exponent 2002, Allen et al 1994). Based on the approximated accelerometer locations, derived center of gravity accelerations from subject 11, whose accelerations and errors approximated those of the total subject mean, were converted to their points of measurement in the previous studies. Based on photographic evidence, the Exponent array was approximated as centered 2.5 cm above the center of gravity of a 50th percentile male subject. This is equivalent to an offset of -2.5 cm in the z-direction for all measurements. The Allen system was more complicated as the sensors did not share a common point of intersection in their lines of measurement unlike the arrays used by this

study and the Exponent study. As a consequence, the Allen sensors do not share a common set of offsets, but each axis of sensitivity has its own. Allen describes the x-axis accelerometer as lying 1.7 cm behind the inion, which is equivalent to a 3.6 cm z-direction offset and a -10.3cm offset in the x-direction. The y-axis accelerometer was described as lying 1.5 cm out from the scalp, directly above the ear. This is approximately -1.4 cm offset in the x-direction, -2.4 cm in the z-direction, and 7.1 cm offset in they-direction. The z-axis was located 2 cm above the vertex, which is approximately an offset of -9.7 cm in the z-direction. Using these offsets, peak acceleration error was calculated as though the measurement locations from these prior studies were used for this study. From this, possible measurement error in these previous studies is calculated.

3.2.2 Equations

Linear accelerations at the cg were calculated based on mouth accelerometer traces based on the transformation below (Naumheim et al 2003). As Y-axis offsets are assumed zero, Y-based terms have been removed from the equations. X and Z are locations of the triaxial block center relative to the center of gravity.

$$a_x^{cm} = a_x^{triax} - (\alpha_y * Z) - \left[(\omega_x * \omega_z * Z) - (\omega_y^2 + \omega_z^2) * X\right] \quad \text{(eq 5)}$$

$$a_y^{cm} = a_y^{triax} - \left[(\alpha_z * X) - (\alpha_x * Z)\right] - \left[(\omega_y * \omega_z * Z) + (\omega_y * \omega_x * X)\right] \quad \text{(eq 6)}$$

$$a_z^{cm} = a_z^{triax} - (-\alpha_y * X) - \left[(\omega_x * \omega_z * X) - (\omega_x^2 + \omega_y^2) * Z\right] \quad \text{(eq 7)}$$

Linear accelerations at the point of measurement for the Exponent and Allen studies are calculated based on the transformation from center of gravity acceleration shown below. In the Exponent equations, X-axis and Y-axis offsets are zero, and have been removed. All terms remain in the Allen equations, and the offsets are unique to each measurement axis due to a lack of a common measurement point.

<u>Exponent (2002)</u>

$$a_x^{Exponent} = a_x^{cm} + (\alpha_y * Z) + [(\omega_x * \omega_z * Z)] \qquad \text{(eq 8)}$$

$$a_y^{Exponent} = a_y^{cm} + [-(\alpha_x * Z)] + [(\omega_y * \omega_z * Z)] \qquad \text{(eq 9)}$$

$$a_z^{Exponent} = a_z^{cm} + [(\omega_x^2 + \omega_y^2) * Z] \qquad \text{(eq 10)}$$

<u>Allen et al (1994)</u>

$$a_x^{Allen} = a_x^{cm} + (\alpha_y * Z^x - \alpha_z * Y^x) + [\omega_x(\omega_y * Y^x + \omega_z * Z^x) - (\omega_y{}^2 + \omega_z{}^2) * X^x] \qquad \text{(eq 11)}$$

$$a_y^{Allen} = a_y^{cm} + (\alpha_z * X^y - \alpha_x * Z^y) + [\omega_y(\omega_z * Z^y + \omega_y * Y^y) - (\omega_x{}^2 + \omega_z{}^2) * Y^y] \qquad \text{(eq 12)}$$

$$a_z^{Allen} = a_z^{cm} + (\alpha_x * Y^z - \alpha_y * X^z) + [\omega_z(\omega_x * X^z + \omega_y * Y^y) - (\omega_x^2 + \omega_y^2) * Z^z] \qquad \text{(eq 13)}$$

3.2.3 Data Analysis

The maximum positive and maximum negative value from each channel plus resultants, angular acceleration, and cg acceleration were recorded from each sample trace. The three samples from each activity were averaged into a subject average for that activity. From this data channel summaries were calculated for mouth and center of gravity accelerations. Peak error was calculated as the difference between peak

accelerations between a mouth axis and its corresponding center of gravity axis, divided by the center of gravity axis. Peak values were simply maxima produced during the six-second collection window. Peak mouth acceleration and center of gravity acceleration would not necessarily occur at the same point in time due to the contribution of the angular terms and the offset of the sensor array from the center of gravity. Angular terms would not necessarily have maximum values at the same time as linear terms. It would be expected that angular acceleration and rate would not have simultaneous peaks due to the nature of angular acceleration calculation – one would expect zero acceleration at maximum positive or negative rates.

To calculate inter-group differences, a matched, two-tailed Student's t-test was used as sample sizes were less than the 30 required for a normalized p-test. Matched tests were used as there was a one-to-one correspondence between Allen, Exponent, mouth, and center of gravity accelerations. T-test scores less than 0.05 were marked as significant. T-test scores between 0.05 and 0.10 were marked as nearly significant.

3.3 RESULTS

3.3.1 Mouth versus CG

In the vast majority of comparisons, the difference between the mean accelerations of the mouth and the center of gravity in the x-, y-, z-directions or the resultant acceleration is statistically significant to the $t < 0.05$ level. Only 25 of the 182 comparisons, or 13.7%, are not significant to this level, and 3 of the remaining 25 are significant to the $t < 0.1$ level. All other comparisons showed statistically significant differences. The majority of insignificantly different comparisons occur for y-axis accelerations, which tend to be small. Negative overall averages had more significantly different results than positive overall averages. This is shown in Tables 12a and 12b.

The mouthpiece accelerometers tended to overestimate the magnitude of the center of gravity accelerometers. Taking an average of peak errors by event, errors in positive maximum values for x-, y-, and z-directions as well as resultants were approximately 20%. Errors in negative maximums in x- and z-directions were particularly high at 43% and 36% respectively. Overall error standard deviations were larger than the overall average error for all but negative-x, making it difficult to identify broad overall trends in error between mouthpiece and center of gravity accelerations. The largest standard deviation is likely driven by the disparate nature of the events. Average error by event, as well as overall average and standard deviation is shown in Table 13.

Table 12a. Overall average positive accelerations (m/s²)

Max +	1	2	3	4	5	6	7	8	9	10	11	12	13
Mouth X	*5.47*	**4.84**	**11.48**	**4.15**	3.31	**3.58**	**8.29**	13.51	**13.45**	22.86	**14.09**	**8.21**	**13.17**
Mouth Y	1.42	1.09	*2.97*	1.22	**13.05**	1.73	*3.16*	*4.42*	**4.46**	7.10	**3.48**	**6.45**	**5.04**
Mouth Z	6.25	**3.69**	**7.93**	**5.78**	**1.39**	3.00	**11.05**	**13.34**	**13.43**	**25.09**	**12.75**	**6.11**	**13.87**
Mouth Resultant	8.43	**7.93**	**23.02**	12.06	**13.38**	**6.97**	**16.65**	**32.84**	**33.43**	**44.43**	**31.19**	**12.87**	**22.48**
CGX	5.25	5.08	10.15	4.44	3.44	3.30	7.02	12.36	12.26	20.32	11.76	6.54	12.32
CGY	1.37	1.20	2.80	0.93	5.28	2.16	3.04	4.04	4.13	6.89	3.21	5.55	4.67
CGZ	6.07	3.61	6.42	4.35	0.82	3.18	10.64	12.62	12.72	21.30	10.95	4.39	12.43
CG Resultant	8.31	7.74	19.68	12.45	6.22	6.40	15.69	30.10	30.60	39.38	28.67	10.06	21.15
Resultant Error	1.4%	2.4%	17.0%	-3.1%	115.2%	9.0%	6.1%	9.1%	9.3%	12.8%	8.8%	27.9%	6.3%

Table 12b. Overall average negative accelerations (m/s²)

Max -	1	2	3	4	5	6	7	8	9	10	11	12	13
Mx	**-2.25**	**-3.13**	**-6.05**	**-10.31**	**-5.74**	**-1.28**	-4.41	**-4.68**	**-4.89**	**-15.59**	**-9.72**	**-6.11**	**-4.50**
My	-1.15	-1.20	-2.44	**-1.18**	**-8.09**	-1.76	-3.27	-3.99	-3.94	**-8.17**	**-3.91**	**-4.75**	**-3.24**
Mz	-6.71	**-5.17**	**-19.37**	**-7.04**	**-2.00**	**-6.11**	-15.18	**-30.24**	**-30.90**	**-38.37**	**-28.51**	**-6.12**	**-18.33**
CGX	-1.98	-1.87	-4.35	-11.65	-2.85	-1.14	-3.72	-3.50	-3.79	-10.17	-5.86	-4.56	-2.18
CGY	-1.12	-1.29	-2.21	-1.06	-3.52	-1.77	-3.15	-3.58	-3.62	-6.75	-3.29	-3.85	-2.71
CGZ	-6.56	-4.92	-16.51	-1.87	-1.25	-5.58	-14.46	-28.15	-28.80	-34.23	-26.70	-3.94	-17.41

All accelerations are in m/s². Values in **bold** are statistically significant to the t < 0.05 level based on a T-Test. Values *underlined in italics* are significant to the t < 0.1 level. Error calculated as (mouth-cg)/cg.

Table 13. Peak error by event.

	1		**2**		**3**		**4**		**5**	
	Max +	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**
Peak X	4.3%	13.8%	-4.7%	67.3%	13.2%	39.2%	-6.6%	-11.6%	-3.7%	101.5%
Peak Y	4.0%	2.3%	-8.8%	-6.8%	6.2%	10.1%	31.4%	11.6%	147.1%	130.0%
Peak Z	2.9%	2.2%	2.1%	5.2%	23.6%	17.3%	33.1%	276.5%	68.4%	59.8%
Peak Resultant	1.4%		2.4%		17.0%		-3.1%		115.2%	
	6		**7**		**8**		**9**		**10**	
	Max +	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**
Peak X	8.7%	12.5%	18.1%	18.6%	9.3%	33.9%	9.8%	29.2%	12.5%	53.3%
Peak Y	-19.6%	-0.6%	3.9%	3.9%	9.5%	11.6%	8.1%	8.9%	3.0%	21.0%
Peak Z	-5.8%	9.5%	3.8%	5.0%	5.7%	7.4%	5.6%	7.3%	17.8%	12.1%
Peak Resultant	9.0%		6.1%		9.1%		9.3%		12.8%	
	11		**12**		**13**		**Overall**		**St Dev**	
	Max +	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**
Peak X	19.8%	65.8%	25.6%	34.1%	6.9%	106.6%	8.7%	43.4%	9.7%	34.6%
Peak Y	8.3%	18.9%	16.3%	23.3%	8.1%	19.6%	16.7%	19.5%	40.9%	34.4%
Peak Z	16.5%	6.8%	39.3%	55.3%	11.6%	5.3%	17.3%	36.1%	20.1%	74.7%
Peak Resultant	8.8%		27.9%		6.3%		17.1%		30.4%	

Error calculated as (mouth-cg)/cg.

There is a slight trend linking peak resultant error with peak resultant angular rate although no obvious trend seems to exist for peak resultant angular acceleration. Angular rate shows a weak trend for a linear curve fit and a reasonable fit for a second order curve fit, with a linear $R^2 = 0.4825$ and a second order $R^2 = 0.6895$. Angular acceleration shows no strong trend for a linear curve fit with a linear $R^2 = 0.0098$. Angular rate affects acceleration via a squared relationship, where as angular acceleration affects linear acceleration via a linear relationship. This is illustrated in equations 8-13. Error appears more strongly associated with the angular rate terms than with the angular acceleration terms. Most angular rate values are clustered between 1 and 2 rad/s whereas angular acceleration values are more distributed. Plots of angular rate and acceleration versus peak error, as well as trend lines, are shown in Figures 24 and 25.

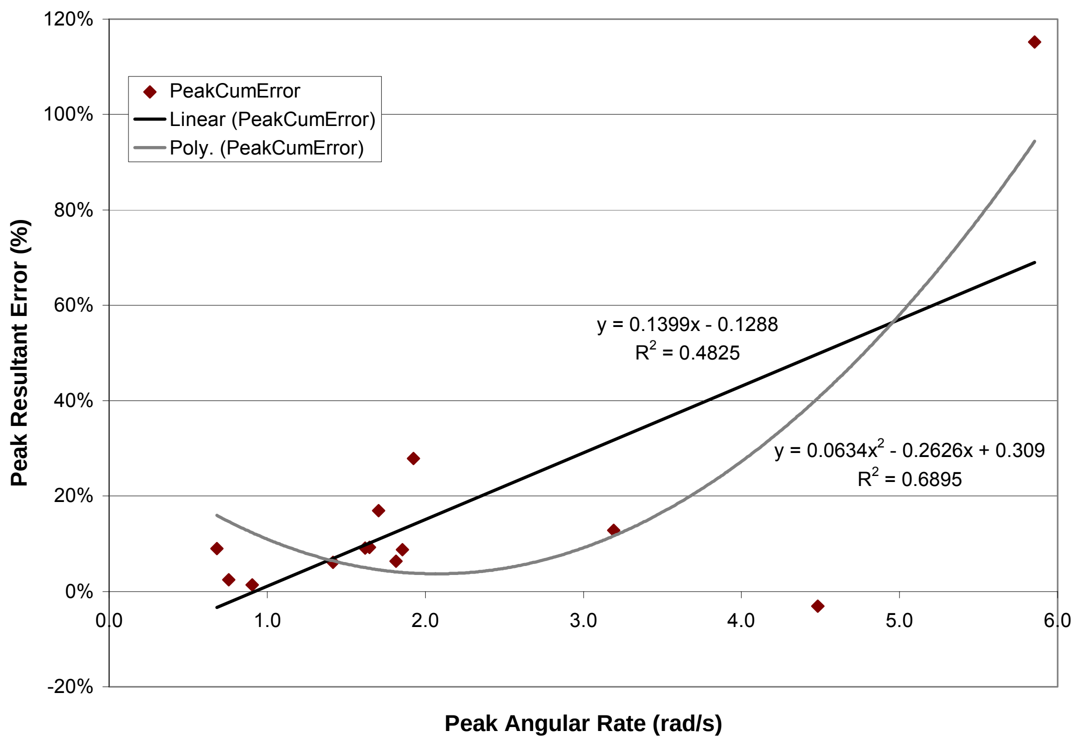

Figure 24. Error in average peak resultant acceleration versus average peak resultant angular rate.

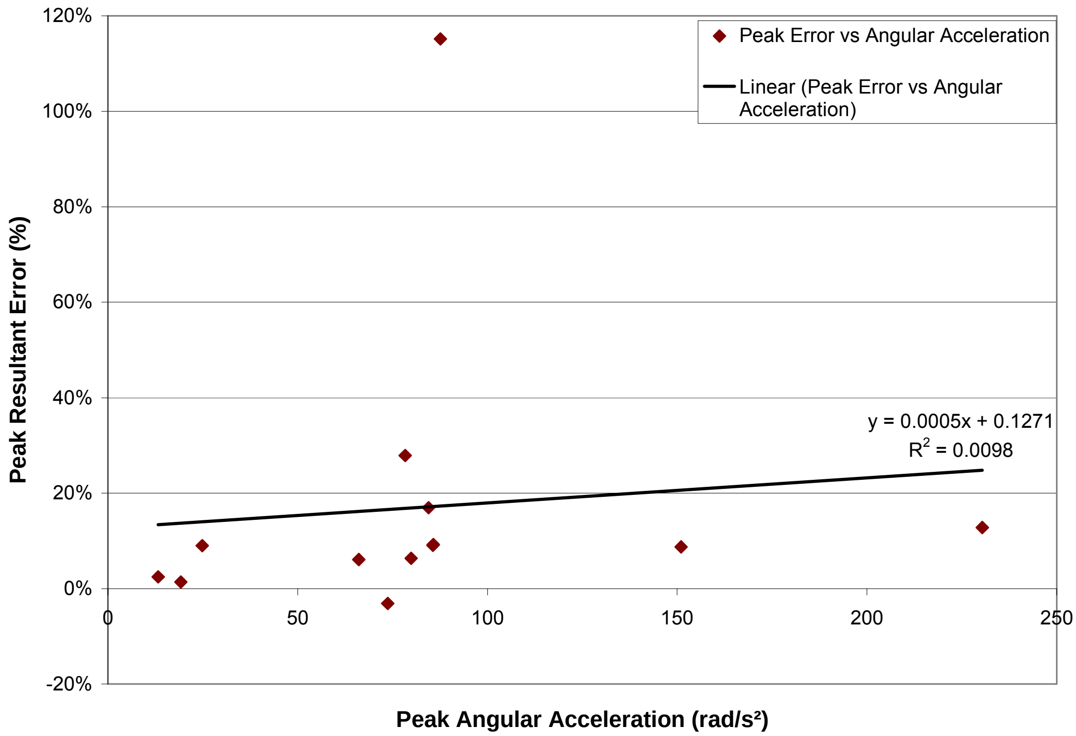

Figure 25. Error in average peak resultant acceleration versus average peak resultant angular acceleration.

3.3.2 Effect on Prior Studies

Based on the data from subject 11, the mouthpiece array, the Allen array, and the Exponent array all show statistically significant angular component influence. The level and frequency of significance are lower than the mouthpiece versus center of gravity comparison above, but this is likely the result of a smaller sample size. The mouth array had approximately the same number of statistically significant comparisons as the Allen and Exponent arrays, despite being overwhelmingly statistically significant in the previous section.

There were statistically significant differences between the center of gravity accelerations and the Allen array accelerations in three of the six activities common to the two studies – Plop Sit, Look Left, and Head Nod/Bob. Partially significant differences –

t < 0.1 – were found for Stair Jump and Chair Stand. Normal Sit did not show statistically significant differences between the center of gravity and the Allen array accelerations.

Table 14 shows the Allen array error for equivalent events from this study. Table 15 compares peak errors between the arrays from this study, Exponent, and Allen. Tables 16a and 16b show the t-test scores by event for the Allen, Exponent, and Mouth arrays as compared to the center of gravity accelerations. Figures 26-31 compare the accelerations from the four locations and show statistically significant differences when applicable.

Table 14. Comparison of Allen and center of gravity accelerations for events common to the two studies. Units in m/s².

	1		2		3		4		5		11	
	Max+	**Max -**	**Max+**	**Max -**	**Max+**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**	**Max +**	**Max -**
CGX	5.67	-3.85	5.20	-9.48	4.58	-14.32	6.03	-15.19	5.14	-6.59	9.82	-2.43
CGY	0.95	-1.41	1.55	-1.59	2.73	-3.27	0.56	-2.01	8.68	-6.92	1.64	-3.61
CGZ	7.02	-7.67	5.37	-1.53	8.87	-13.63	5.56	-4.08	1.18	-2.18	10.43	-21.82
CG Result	9.07		10.35		18.15		15.62		8.88		23.53	
AllenX	5.69	-3.65	5.19	-9.44	4.63	-15.24	*5.68*	*-13.02*	*7.09*	**-4.39**	8.56	-4.39
AllenY	0.97	-1.44	1.54	-1.61	2.74	-3.25	0.57	-2.05	**5.63**	*-7.92*	1.68	*-3.23*
AllenZ	7.03	-7.64	5.39	-1.50	8.92	**-13.56**	**8.03**	-3.82	1.23	*-2.12*	10.44	-21.75
AllenR	*9.00*		10.35		18.99		*13.35*		9.66		23.25	
Peak X Error	0.4%	-5.2%	-0.4%	-0.4%	1.0%	6.4%	-5.8%	-14.3%	38.0%	-33.4%	-12.8%	80.5%
Peak Y Error	2.4%	2.1%	-0.8%	1.3%	0.1%	-0.7%	0.6%	2.1%	-35.1%	14.5%	2.1%	-10.7%
Peak Z Error	0.2%	-0.4%	0.5%	-2.4%	0.6%	-0.5%	44.3%	-6.4%	3.5%	-2.7%	0.1%	-0.3%
Peak R Error	-0.7%		0.0%		4.6%		-14.5%		8.8%		-1.2%	

Highlighted values correspond to significant values from Tables 16a and 16b. Error calculated as (Allen-CG)/CG.

Table 15. Comparison of peak error between this study, Exponent and Allen. All values in %.

Event	1	2	3	4	5	6	7	8	9	10	11	12	13
Mouth vs CG - X+	2.9%	2.1%	2.2%	-12.3%	1.8%	5.6%	25.9%	8.7%	8.7%	-7.6%	17.6%	22.5%	11.5%
Mouth vs CG - X-	2.8%	0.0%	17.9%	-19.2%	96.4%	7.9%	49.9%	23.4%	23.4%	122.2%	212.5%	16.3%	-0.4%
Mouth vs CG - Y+	13.3%	-1.4%	6.6%	86.6%	193.1%	4.4%	-8.4%	21.5%	21.5%	-10.4%	-0.4%	41.3%	34.9%
Mouth vs CG - Y-	9.4%	0.0%	15.7%	33.7%	145.1%	-2.4%	0.8%	10.1%	10.1%	16.5%	25.5%	33.1%	10.6%
Mouth vs CG - Z+	7.0%	9.6%	7.5%	64.4%	93.5%	12.6%	3.4%	7.6%	7.6%	47.4%	22.6%	144.7%	8.6%
Mouth vs CG - Z-	6.2%	0.0%	-17.6%	198.1%	82.5%	-1.7%	-1.1%	0.8%	0.8%	6.0%	15.9%	77.8%	6.0%
Mouth vs CG - Res	2.6%	2.0%	3.7%	-3.1%	189.0%	2.8%	1.2%	3.5%	3.5%	-1.2%	18.6%	40.4%	6.9%
Exp vs CG - X+	0.2%	2.0%	1.2%	12.6%	-1.1%	1.9%	-6.2%	-1.4%	-1.4%	30.1%	9.8%	13.4%	16.8%
Exp vs CG - X-	4.0%	0.4%	3.7%	8.6%	-13.7%	6.9%	-5.6%	-1.8%	-1.8%	47.7%	40.9%	9.5%	60.8%
Exp vs CG - Y+	1.2%	-2.2%	0.8%	-2.3%	-11.0%	3.7%	6.0%	8.3%	8.3%	8.9%	2.5%	-30.9%	5.5%
Exp vs CG - Y-	3.9%	1.3%	-1.4%	2.2%	-5.3%	0.3%	1.1%	1.9%	1.9%	-10.4%	-11.6%	-7.2%	-9.4%
Exp vs CG - Z+	0.0%	0.1%	0.2%	10.7%	0.9%	0.0%	0.0%	0.1%	0.1%	0.1%	0.0%	0.4%	0.1%
Exp vs CG - Z-	-0.4%	-2.4%	-0.5%	-6.4%	-2.7%	0.0%	-0.1%	0.0%	0.0%	-0.7%	-0.3%	-3.7%	-1.3%
Exp vs CG - Res	0.5%	0.1%	1.0%	8.2%	-9.4%	2.5%	0.5%	0.3%	0.3%	6.7%	1.7%	-0.2%	1.2%
Allen vs CG - X+	0.4%	-0.4%	1.0%	-5.8%	38.0%	2.3%	13.1%	4.5%	4.5%	-32.8%	-12.8%	2.6%	-3.8%
Allen vs CG - X-	-5.2%	-0.4%	6.4%	-14.3%	-33.4%	-1.3%	21.4%	3.7%	3.7%	37.3%	80.5%	-0.4%	-22.9%
Allen vs CG - Y+	2.4%	-0.8%	0.1%	0.6%	-35.1%	8.2%	5.5%	9.9%	9.9%	5.8%	2.1%	-24.0%	0.6%
Allen vs CG - Y-	2.1%	1.3%	-0.7%	2.1%	14.5%	1.6%	2.0%	1.4%	1.4%	-7.8%	-10.7%	-0.7%	-4.6%
Allen vs CG - Z+	0.2%	0.5%	0.6%	44.3%	3.5%	0.2%	0.1%	0.3%	0.3%	0.5%	0.1%	1.4%	0.3%
Allen vs CG - Z-	-0.4%	-2.4%	-0.5%	-6.4%	-2.7%	0.0%	-0.1%	0.0%	0.0%	-0.7%	-0.3%	-3.7%	-1.3%
Allen vs CG - Res	-0.7%	0.0%	4.6%	-14.5%	8.8%	-0.3%	0.2%	-0.3%	-0.3%	-9.2%	-1.2%	-8.3%	-2.4%

Error calculated as (Array-CG)/CG.

Table 16a. T-scores for peak positive accelerations by event.

Event	1	2	3	4	5	6	7	8	9	10	11	12	13
Mouth vs CG - X	0.484	0.524	0.307	**0.044**	**0.022**	**0.005**	**0.028**	*0.052*	0.212	0.699	0.336	0.172	0.365
Mouth vs CG - Y	**0.021**	0.785	0.233	*0.065*	**0.017**	0.676	0.324	*0.058*	0.262	0.300	0.983	*0.068*	0.267
Mouth vs CG - Z	**0.047**	0.076	0.582	**0.011**	0.146	0.155	0.118	0.250	*0.097*	0.229	0.407	0.172	0.161
Mouth vs CG - R	0.196	0.553	0.511	0.548	**0.016**	**0.040**	0.276	**0.028**	0.458	0.910	0.227	0.197	0.437
Exp vs CG - X	0.895	0.233	0.118	*0.054*	*0.076*	0.769	0.066	0.993	0.196	**0.018**	0.353	0.233	0.413
Exp vs CG - Y	0.412	**0.002**	0.494	0.688	**0.006**	0.123	0.111	0.829	**0.033**	**0.002**	0.904	*0.067*	**0.021**
Exp vs CG - Z	0.248	0.191	0.336	*0.055*	0.178	0.160	*0.089*	0.234	*0.054*	*0.094*	0.351	0.353	*0.097*
Exp vs CG - R	0.251	0.881	0.786	**0.016**	**0.011**	0.677	**0.036**	0.874	0.675	**0.042**	0.397	0.970	0.347
Allen vs CG - X	0.830	0.883	0.387	*0.061*	*0.082*	**0.036**	**0.041**	0.109	0.191	**0.018**	0.411	0.349	0.529
Allen vs CG - Y	0.407	0.340	0.852	0.810	**0.020**	0.130	**0.039**	0.592	**0.031**	*0.051*	0.880	*0.074*	0.270
Allen vs CG - Z	0.248	0.191	0.336	**0.042**	0.178	0.160	*0.089*	0.234	0.141	*0.094*	0.351	0.353	*0.097*
Allen vs CG - R	*0.089*	0.989	0.115	*0.053*	0.419	0.851	0.625	**0.034**	0.740	**0.038**	0.456	0.331	0.321

Table 16b. T-scores for peak negative accelerations by event. *

Event	1	2	3	4	5	6	7	8	9	10	11	12	13
Mouth vs CG - X	0.749	0.891	0.278	**0.025**	**0.003**	*0.095*	*0.074*	0.284	**0.049**	**0.010**	0.306	0.349	0.988
Mouth vs CG - Y	0.214	0.151	0.519	*0.061*	**0.000**	*0.086*	0.862	0.945	0.459	**0.019**	*0.095*	**0.034**	0.281
Mouth vs CG - Z	0.142	0.487	0.218	**0.002**	**0.023**	0.439	0.663	*0.058*	0.877	0.532	0.203	*0.100*	0.484
Exp vs CG - X	0.260	0.655	0.337	**0.020**	**0.003**	0.751	*0.084*	0.313	0.615	0.253	0.436	0.356	0.337
Exp vs CG - Y	0.113	0.353	0.680	0.673	**0.023**	0.181	0.516	0.608	0.374	**0.016**	0.175	0.424	*0.073*
Exp vs CG - Z	0.188	0.246	**0.042**	0.177	*0.092*	0.162	*0.089*	0.333	0.184	0.267	0.355	**0.013**	0.335
Allen vs CG - X	0.304	0.767	0.304	*0.061*	**0.028**	0.344	0.103	0.349	0.371	**0.007**	0.392	0.959	0.539
Allen vs CG - Y	0.306	0.462	0.531	0.632	*0.071*	**0.042**	0.338	0.336	0.755	**0.010**	*0.086*	0.941	**0.032**
Allen vs CG - Z	0.188	0.246	**0.043**	0.180	*0.092*	0.162	*0.089*	0.165	0.184	0.267	0.355	**0.013**	0.332

* Resultant accelerations cannot be negative.
Events 1-5 and 11 in gray to highlight common events with Allen et al 1994.

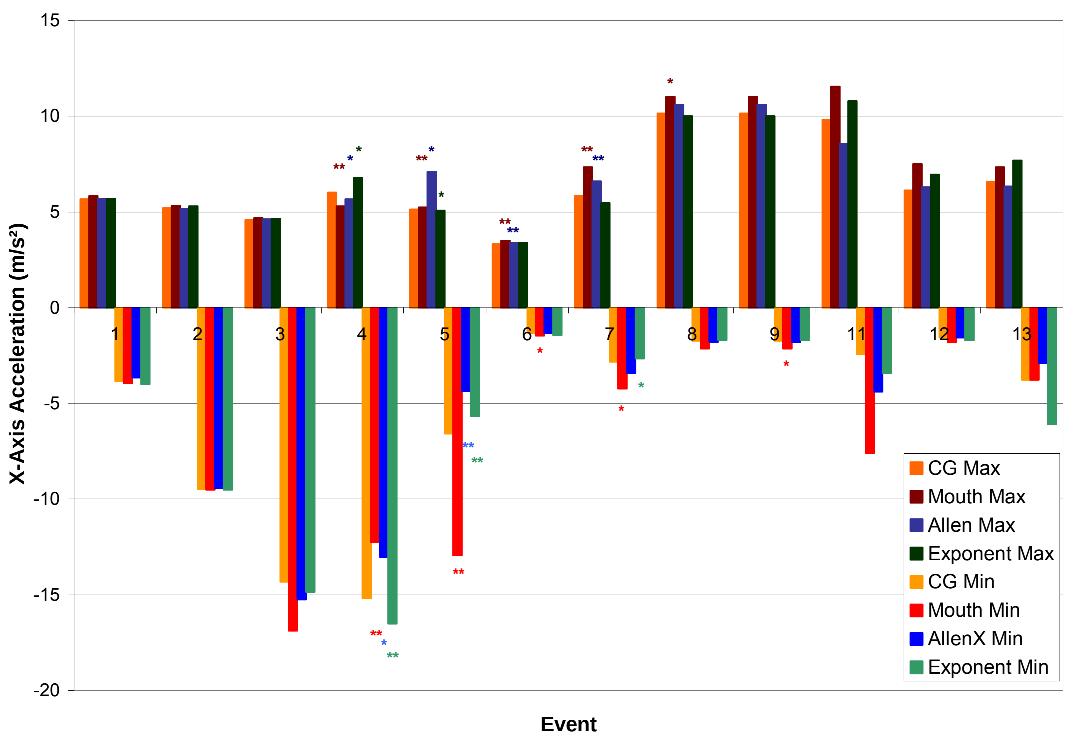

Figure 26. Comparison of x-axis accelerations between center of gravity, mouth array, Allen array, and Exponent array. Significance to t < 0.1 shown by "*". Significance to t < 0.05 shown by "".**

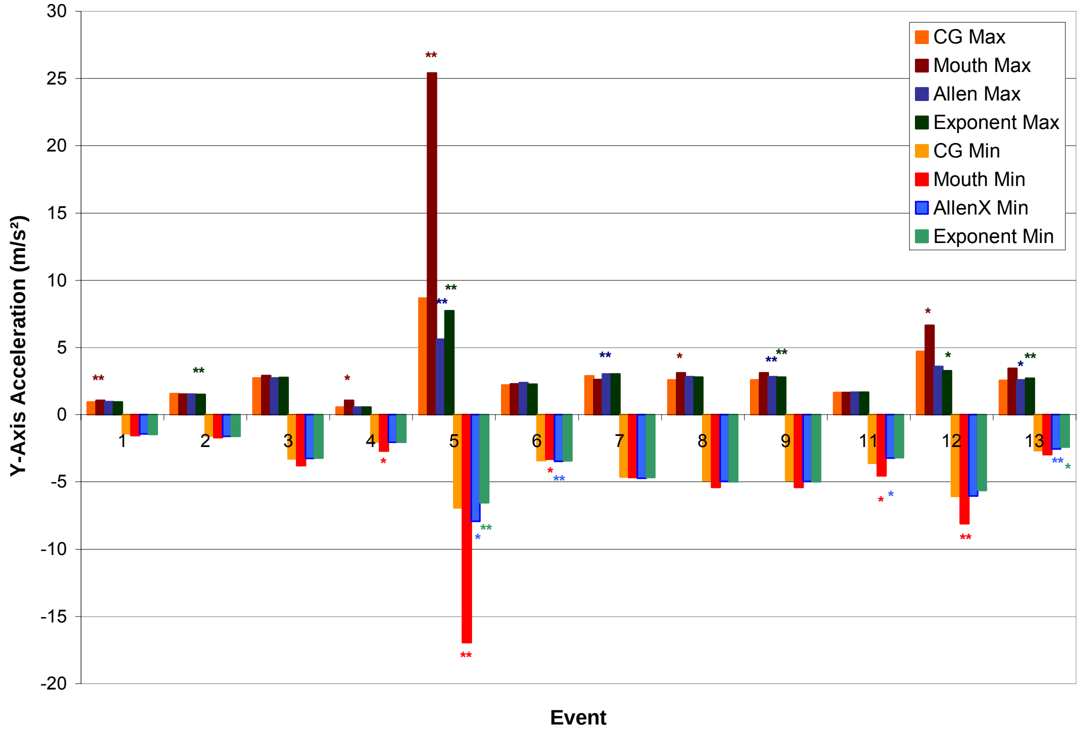

Figure 27. Comparison of y-axis accelerations between center of gravity, mouth array, Allen array, and Exponent array. Significance to t < 0.1 shown by "*". Significance to t < 0.05 shown by "".**

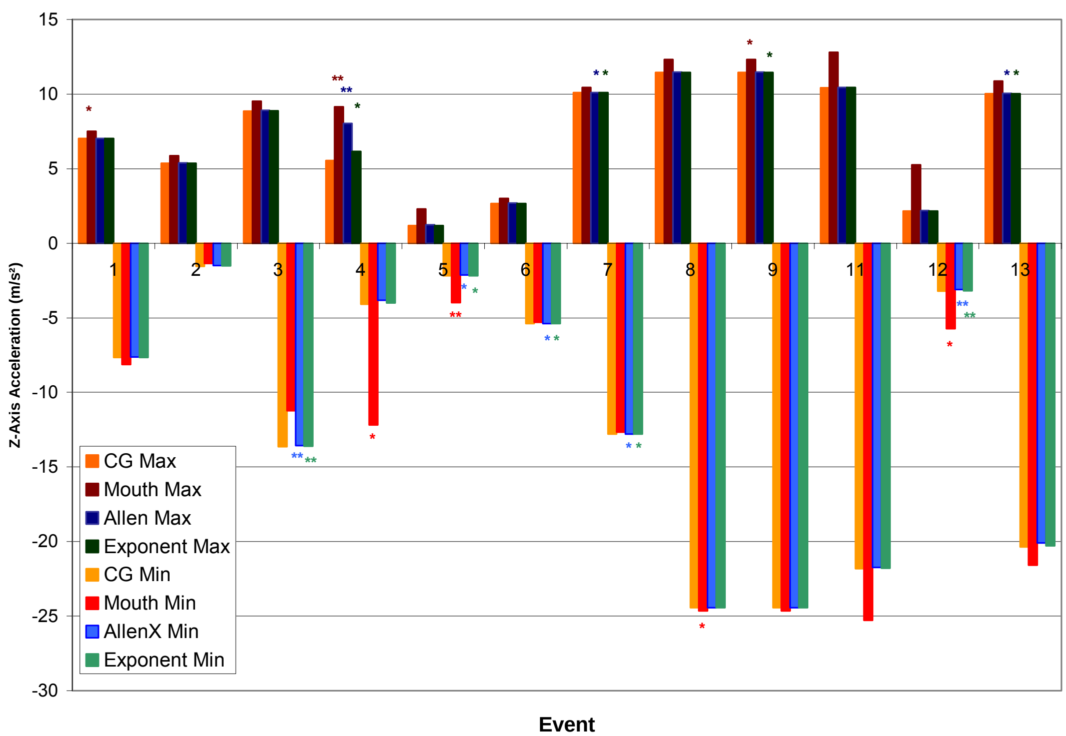

Figure 28. Comparison of z-axis accelerations between center of gravity, mouth array, Allen array, and Exponent array. Significance to t < 0.1 shown by "*". Significance to t < 0.05 shown by "".**

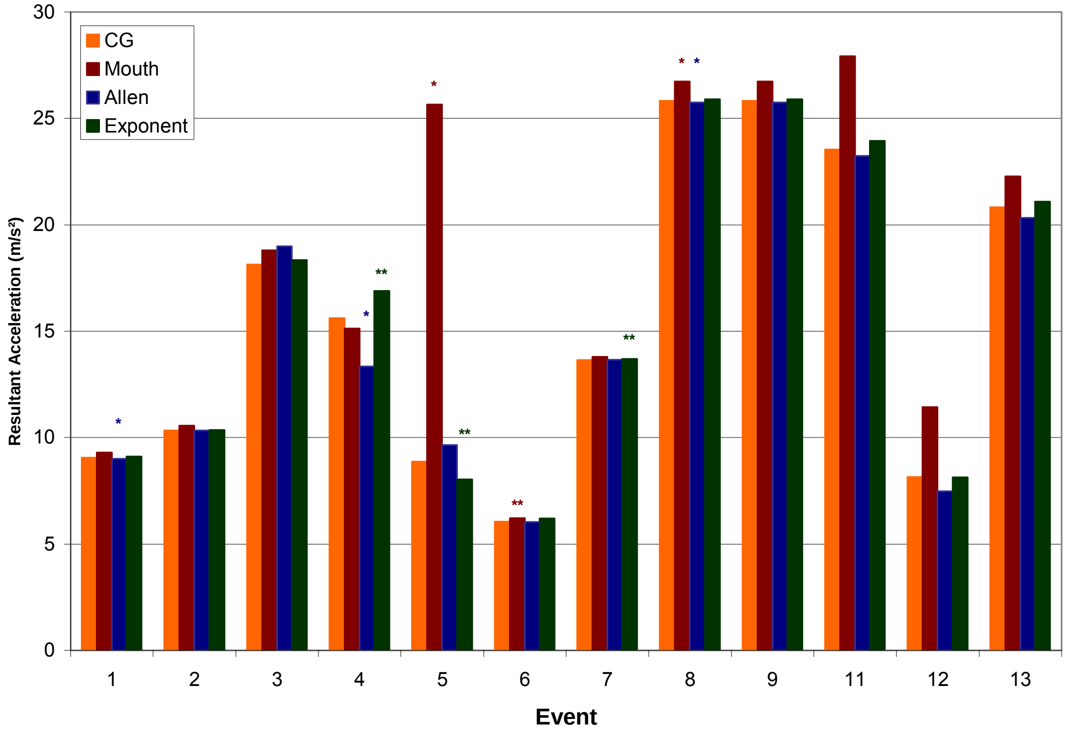

Figure 29. Comparison of resultant accelerations between center of gravity, mouth array, Allen array, and Exponent array. Significance to t < 0.1 shown by "*". Significance to t < 0.05 shown by "".**

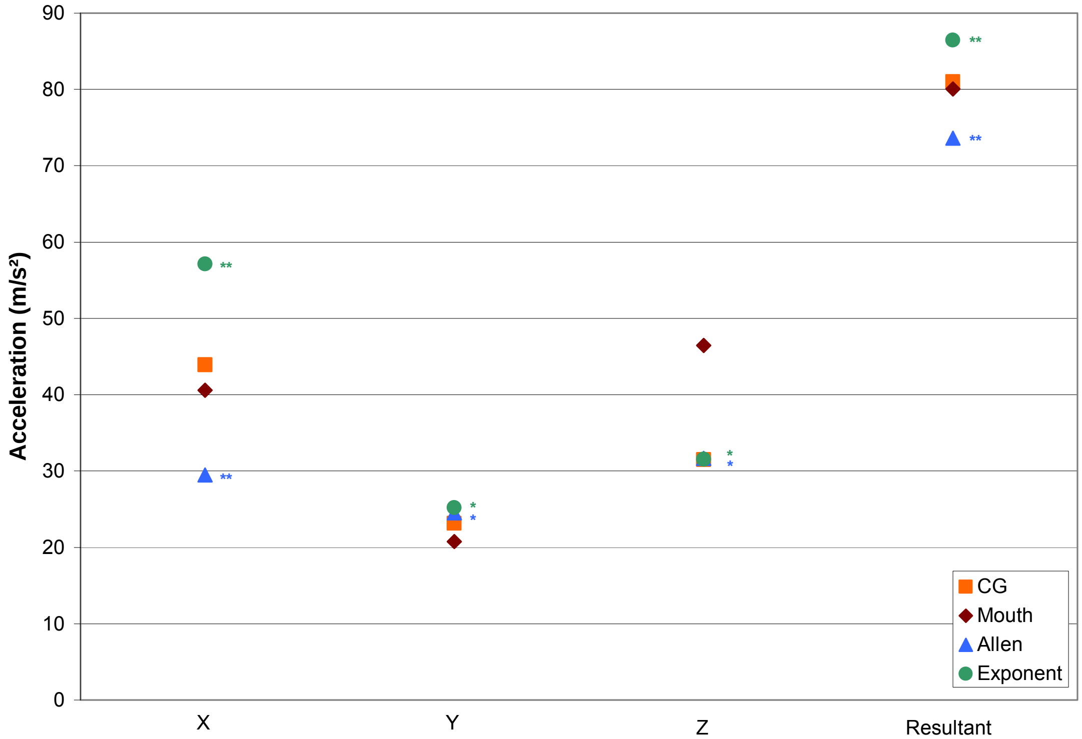

Figure 30. Comparison of accelerations between center of gravity, mouth array, Allen array, and Exponent array for positive directions of Event 10. Significance to t < 0.05 shown by "".**

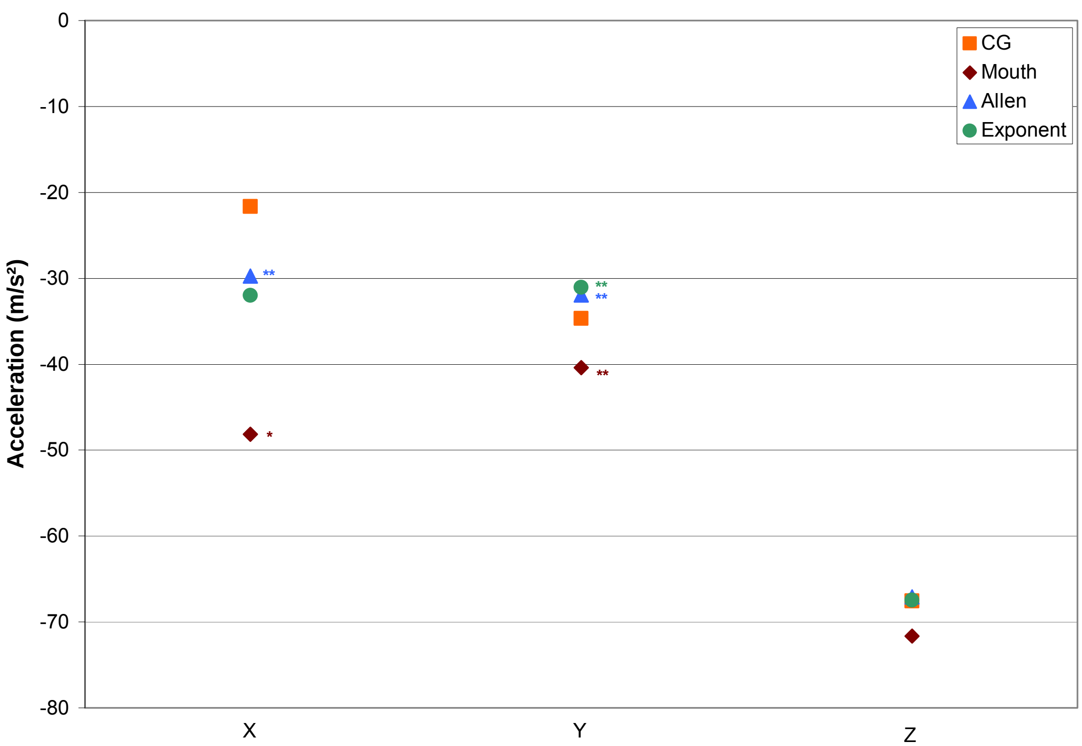

Figure 31. Comparison of accelerations between center of gravity, mouth array, Allen array, and Exponent array for negative directions of Event 10. Significance to t < 0.05 shown by "".**

3.4 DISCUSSION

3.4.1 Mouth versus CG

While mean error between mouth and cg was large, standard deviations were larger. This effect is most likely caused by the wide variety of error from event to event stemming from the disparate nature of the events themselves. The complex interrelationship of array offsets, angular terms and dominant linear directions led to the very different errors from event to event seen in these results. Despite this, trends appear. The mouthpiece usually overestimated the center of gravity accelerations, likely because rotational motion of the head induced larger accelerations at the periphery of the head than at the center of gravity. A notable exception is Head Nod, where the mouthpiece underestimated x-axis accelerations. These errors were relatively small in the less intense activities, but even in Chair Sit there is a 67% error on the negative-x acceleration. Mouthpiece accelerations were often 20% or greater different than the corresponding center of gravity acceleration.

This difference between mouthpiece and center of gravity acceleration, combined with nearly unanimous statistical significance of differences, demonstrate that the effects of angular rate and acceleration cannot be neglected – even for everyday activities. It is particularly notable that this result also applies to events 6 and 7 – walking and running – as these are commonly measured activities in the literature which typically neglect angular rate or acceleration measurement. This study shows that accelerations measured by triaxial arrays mounted on the head may be significantly different than the accelerations at the center of gravity of the head.

All of the arrays tended to overestimate or underestimate center of gravity accelerations with the effect most pronounced in events with significant rotational components, such as Head Nod or Maximum Vertical Leap. The tendency of the mouthpiece to overestimate the magnitude of center of gravity accelerations appears to be linked to the placement of the accelerometers below and in front of the center of gravity. The Exponent and Allen arrays behaved similarly with the Exponent array somewhat more accurate. Unlike the mouthpiece, errors were both in under- and overestimation of center of gravity acceleration. In general, the closer an array is to the center of gravity and the fewer dimensions in which it is offset, the more accurate it is. Error, as expected, increased with angular rate and acceleration.

The quirk of an offset accelerometer underestimating center of gravity accelerations is non-intuitive. The interpretation is that the offset accelerometer location is rotating counter to the linear direction of head acceleration. Using Head Nod with the mouthpiece array as an example, smaller x-axis accelerations can be produced if angular rates are small and angular accelerations act counter to the linear acceleration direction. In this case, maximum acceleration in x occurs after the initial nod and right as the return action begins. At this instant, angular rates are near zero and there is a large positive angular acceleration about the y-axis. Based on equation 5, a positive angular term multiplied by a negative offset is subtracted from a negative acceleration. The net result is an acceleration added to the negative linear acceleration, reducing its magnitude.

Although difficult to achieve due to the location of the center of gravity, a triaxial array with a point of intersection of the lines of sensitivity lying within the center of gravity would theoretically not show the over- or underestimation errors seen in this and

prior studies. The chief difficulty would be guaranteeing the orientation of the accelerometers relative to each other and the center of gravity, as the center of gravity lies below the eyes and essentially on top of the ears.

It is important to remember when noting the relative fit of the trend lines to the angular rate and acceleration data that correlation does not mean causation. The peak resultant angular rates correlate better with the peak resultant errors, but this does not necessarily mean they cause them. As in the above example, with Head Nod, error appears to be driven almost solely by angular acceleration with almost zero contribution from angular rate. With the disparate nature of the events in this study, error is most likely shared between rates and accelerations with the dominant contributor varying from event to event. The better correlation with angular rate may have as much to do with the tendency of peak resultant angular rates to be around 1.5 rad/s as it does with any material factor. Simply, that the angular rate data is less scattered than the angular accelerations does not mean they are more a cause of error.

It is important to keep in mind calculation method when comparing peak accelerations. This study simply found trace maxima when determining peak acceleration terms. As such, the peak mouth resultant acceleration could occur at a wholly different time than the peak center of gravity resultant acceleration. This is partially because angular terms often lag or lead linear acceleration terms and the maxima of angular components of offset arrays do not necessarily occur in sync with the linear maxima. With the cyclic events, such as walking or running, a peak mouth value may occur on a different step than the center of gravity peak value. Peak center of gravity values were not found by finding the peak mouth maxima and converting at that point in

time, rather center of gravity accelerations were calculated at each time point and maxima found independently. This is illustrated in Figure 32.

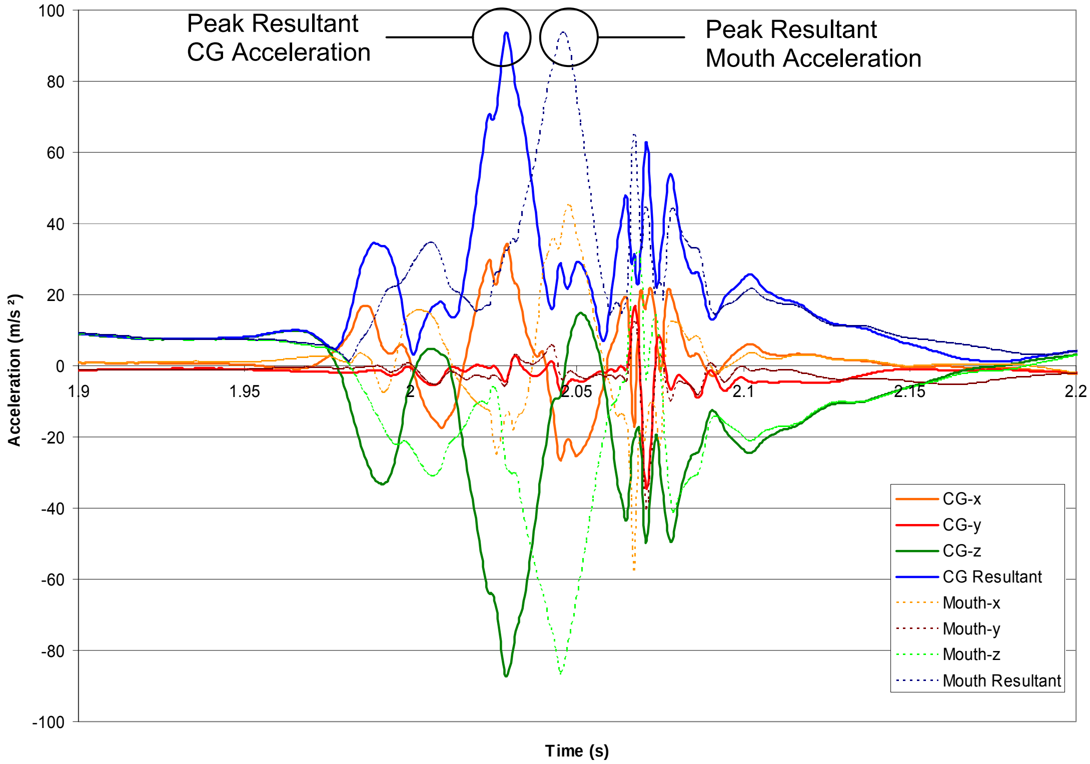

Figure 32. Timing of peak resultant acceleration for mouth and cg locations.

3.4.2 Effect on Prior Studies

Although the small sample size makes conclusions less clear, it appears that the Exponent and Allen arrays are just as affected by angular rate and acceleration as the Mouth array. That the Exponent array is the least affected by the angular terms is unsurprising, as its point of measurement lies closest to the center of gravity of the head. Despite this, however, it shows nearly as many statistically significant differences in a matched, two-tailed T-test as the Allen and Mouth arrays.

It is difficult to revisit the Exponent results and attempt to apply correction factors, as without any common activities it is difficult to estimate the magnitude and

direction of the angular terms involved. It is somewhat easier to attempt to correct some of the Allen results, however, as there were common activities. This study has analogues to Allen's Plop in chair, Hop off step, Sit down, Stand up, and Head bob. Look Left from this study was conducted more quickly than Allen's Look left and does not readily compare. As is previously shown in Table 14, Head Nod and Stair Jump were the most susceptible to component error with the Allen array. This indicates that Allen's Head bob and Hop off step most likely exhibit similar error. Table 17 shows the results from comparable Allen activities with center of gravity correction applied. The results for Stand up, Sit down, and Plop in chair do not change appreciably, but both the x-axis and vector terms are substantially affected for Hop off step and Head bob – approximately 2 G for x-axis Hop off step.

Table 17. Maximum Allen results versus maximum center of gravity corrected values.

	Allen		Corrected	
	-x (g)	**Vector (g)**	**-x (g)**	**Resultant (g)**
Stand up	1.5	1.9	1.6	1.9
Head bob	1.9	2.2	2.5	2.8
Sit down	1.6	2.5	1.5	2.5
Hop off step	4.5	8.1	2.5	7.2
Plop in chair	5.6	10.1	5.3	9.9

Allen et al recorded data only in the negative x and positive-z directions. For purposes of correction, negative-x and z-axis linear acceleration error terms have been used.

While corrections of 0.5-2 G do not seem of clinical relevance, it should be noted that whiplash cases are reported at the 4-5 G level and below; meaning this amount of error could represent a 10-50% change in injury thresholds (Allen et al 1994). This study demonstrates that measurement of daily activities conducted without correction for angular terms can show error of up to 80% for component or resultant accelerations. All accelerations from such studies should be taken with a grain of salt.

3.5 CONCLUSION

Accelerometer array locations relative to the center of gravity of the head from this study and two prior studies were determined and used to calculate error terms relative to the center of gravity accelerations. Mouth versus center of gravity comparisons were based on 700 samples across 13 activities and 18 subjects. Comparisons with Exponent and Allen arrays were based on 39 samples across 13 activities using subject 11.

The mouth accelerations were almost unanimously statistically significantly different than center of gravity accelerations. Mouth accelerations typically overestimated center of gravity accelerations. Error increased with increasing angular rates and accelerations.

The Exponent and Allen arrays were similarly affected by angular terms, although to a lesser extent than the mouth array. The Exponent and Allen arrays both under- and overestimated center of gravity accelerations. The Exponent and Allen arrays show similar frequency of statistically significant differences with center of gravity acceleration as the mouth array across the same data set. Based on common events, four Allen results have been adjusted to center of gravity accelerations. Correction magnitudes were as large as 2 G.

It has been shown that angular terms have a significant effect on this and prior studies of daily accelerations. It is not sufficient to conclude that accelerations measured away from the center of gravity of the head adequately represent head accelerations.

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Graduate Research Assistant
- Advisor – Dr. Stefan Duma
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- Perform data collection during practices and games for multi-institution study on head impacts in football
- Design and build head accelerometer array
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National Science Foundation Research Experience for Undergraduates at University of Kansas May-August 2002
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Intern – Internet and Shared Services

- Create client tutorials and researched server applications.
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- Work directly with site managers to maintain, troubleshoot, and redesign KitchenAid and Whirlpool branded websites.

SKILLS

- Multiple-channel data acquisition
- CAD design utilizing Pro/Engineer and I-DEAS
- Dynamic simulation utilizing MADYMO and LS-DYNA
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RESEARCH INTERESTS

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- Transportation Injury Biomechanics
- Injury Criteria Development

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Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, Greenwald RM, Chu JJ, Crisco JJ. (2005) Analysis of Real-time Head Accelerations in Collegiate Football Players. Clinical Journal of Sports Medicine, 15:3-8.

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- President, Michigan Tech Fencing Club (2000-2002)
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PROFESSIONAL SOCIETIES

- Society of Automotive Engineers (SAE) Member (Student)
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- ASTM F24 Committee Member
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- Biomedical Engineering Society (BMES) – MTU Chapter (2000-2003)
- United States Fencing Association, *Senior Competitive Member* (1999-2005)
- Michigan Tech Fencing Club (1999-2003)
- Michigan Tech Intramurals, *Broomball* (1999-2003)