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Muscle activation, elbow angle and wrist guard effects on wrist and elbow response following simulated forward falls

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Muscle Activation, Elbow Angle and Wrist Guard Effects on Wrist and Elbow Response Following Simulated Forward Falls

by

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Abstract

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The purposes of this study were to: i) determine the acceleration response of the forearm at the wrist and elbow; ii) determine the efficacy of wrist guards by measuring the impact outcome at the wrist and elbow, and; iii) determine the effects of muscle activation and elbow angles on the wrist and elbow acceleration characteristics.

A seated human pendulum was designed to simulate a forward fall and produce an impact to the hands of 28 subjects. Two surface accelerometers measured the wrist and elbow acceleration characteristics in response to four muscle activation levels two elbow angles and a wrist guard.

The results suggest that wrist guards are capable of absorbing or redirecting the initial impact force. Furthermore, the increase in muscle activation level and the natural elbow angle, disabled the segments ability to attenuate the shock wave initiated at the hand as measured by the acceleration response at the elbow.

Dedication

This thesis is dedicated to my parents, Donna and Lee Burkhart
your support, constant motivation and unconditional love have made this possible. I cannot express my love and gratitude enough.

Acknowledgements

I would first like to thank my Advisor Dr. David Andrews, for his commitment and dedication to this paper. Your patience, support and willingness to teach throughout this journey will never go unappreciated. The passion you show towards your work has been a constant inspiration, thank you for everything.

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Finally, I would like to thank my brother and sister-in-law, Mathew and Carly and my.... Amanda. To my brother and sister-in-law, you have both been a constant outlet of support, thank you for standing by me throughout this journey. Amanda, thank you for your constant encouragement, and for helping me to see the best in even the most difficult situations. I hope you know how much you mean to me as this would not have been possible without you.

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Glossary of Terms

ART (Available reaction time) – Period beginning from the initiation of a fall until the instant of impact.

AS_{axial} – Axial acceleration slope measured between 30 % and 70 % of the peak acceleration.

AS_{off} – Off-axis acceleration slope measured between 30 % and 70 % of the peak acceleration.

b (Damping coefficient) – Represents the magnitude of oscillation resistance in a mechanical system.

BMA (Bone mounted accelerometer) – Direct attachment of an accelerometer to the bone.

BW (Body weight) – The measure of the force of gravity acting on the body measured in Newtons.

CM (Centre of mass) – The point on a body or segment that moves as though all the mass were concentrated there and all external forces were applied there.

GRF (Ground reaction force) – A force that is equal in magnitude and opposite in direction to the force that the body exerts onto a supporting surface through the hand.

HAT (Head arm torso complex) – A lumped sum mass representing the upper body segment.

IRF (impact reaction force) – A force that is equal in magnitude and opposite in direction to the force that the body exerts onto a supporting surface through the hand.

k (Stiffness coefficient) – Represents the magnitude of stiffness in a mechanical system.

MVE (Maximal voluntary exertion) – Maximal force output a muscle is capable of achieving.

PA_{axial} (Peak axial acceleration) – Maximal acceleration measured parallel with the long axis of the segment.

PA_{off} (Peak off-axis acceleration) – Maximal acceleration measured at a right angle to the long axis of the segment

SMA (Skin mounted accelerometer) – An accelerometer attached directly to the skin.

TPA_{axial} (Time to peak axial acceleration) – Time period between the impact and the maximal axial acceleration.

TPA_{off} (Time to peak off-axis acceleration) – Time period between the impact and the maximal off-axis acceleration.

WSIB (Workplace Safety and Insurance Board) - An organization that promotes workplace health and safety, and provides a workers compensation system for the employers and workers of Ontario.

WM (Wobbling mass) – Muscles and soft tissue that are attached to the underlying bony structures.

Chapter I
Introduction

Slip, trip and fall injuries are a continuing societal problem as the associated pathologies can be extremely detrimental and often costly. Direct occupational costs of injuries associated with slips, trips and falls is estimated to exceed US $6 billion a year (Chang, 2002). Falls can occur in any direction (i.e. to the side, backwards, or forwards), and although backwards falls produce greater hand forces (Tan, Eng, Robinovitch & Warnick, 2006), Nevitt & Cummings (1993) have found that forward falls occur much more frequently. When falling forward, it is common for an individual to defend against the potentially detrimental effects to the head and torso by extending the arm to accept the initial impact and arrest the forward momentum of the body (Hsiao & Robinovitch, 1998). However, while trying to avoid head and torso injuries, the upper extremity can experience a significant impact, potentially leading to a variety of injuries including distal radius fractures (Frykman, 1967; Chiu & Robinovitch, 1998), proximal ulna fractures (Doornberg & Ring, 2006), fractures to the scaphoid (Leslie & Dickson, 1981), and fractures to the distal humerus (Houshian, Mehdi, & Larsen, 2001).

Upwards of 55% of the people in the general work force (Chang, 2002), 53% of the elderly (Blake et al, 1988) and 80.4 % of those who participate in-line skating (Ellis, Kierluf, & Klassen, 1995) will experience a forward fall, resulting in some form of upper extremity injury. Although there is a general understanding of how, and where these falls are occurring, there remains a need to move beyond measures which act to prevent the fall altogether, to a more complete understanding how the limb reacts upon impact. A clear understanding of both the external biomechanical demands applied at impact and the biological capacity of the tissues is crucial in order to prevent injuries related to impacts. Strategies aimed at decreasing impact effects must be successful in dissipating impact energy and/or diverting the force away from the susceptible bony structures.

The effectiveness of wrist guards for decreasing force at impact has been researched extensively. The rigid volar splint in many wrist guards is designed to divert the impact away from the carpals of the wrist, and absorb the majority of the impact energy. However, wrist guard use remains a topic of debate in the literature, as many feel that the conventional volar splint design is successful in preventing a wrist injury such as a distal radius fracture (Schielbler et al., 1996; Greenwald, Janes, Swanson & McDonald, 1998), while others believe the efficacy of this equipment is less than optimal (Cheng, Rajaratnam, Raskin, Richard & Axelrod, 1995; Giacobetti, Sharkey, Bos-Giacobetti, Hume & Taras, 1997).

Similar to protective equipment, the body itself is also capable of implementing force reducing mechanisms in its ability to actively (e.g. by changing joint angles and muscle activation levels), and passively (e.g. by displacement and deformation of soft tissues) attenuate impact force outcomes. Recent work has considered the effects of different arm orientations (Chou et al., 2001; DeGoede, Ashton-Miller, Schultz & Alexander 2002), and the ability to decrease the magnitude of ground reaction force peaks experienced at the hand. Degoede & Ashton-Miller (2002) found that by flexing the elbows during impact resulting from a forward fall, an individual is capable of actively lowering the impact force peaks by approximately 60 %. These posture effects have led to the implementation of fall interventions (Lo, McCabe, DeGoede, Okuizumi, & Ashton-Miller, 2003), and have proven relatively successful at reducing ground reaction forces overall.

Soft tissues, especially muscle, are capable of acting as passive shock attenuators in living subjects, thereby reducing the effects of applied impacts. Changes in muscle activation, potentially leading to changes in muscular stiffness, also play a key role in the attenuation of force and the dissipation of energy within and between body segments. Changes in muscular stiffness as a result of increased activation are achieved through the interaction of a greater number of cross bridges (Mijailovich, Fredberg & Butler, 1996; Lee, Rogers & Granata, 2006). Laquinta, Licata, & Soechting (1982) and Osu & Gomi (1999), studying the upper extremity, were both able to provide evidence that as the forearm and arm musculature was activated to greater levels, the stiffness about the elbow joint proportionally increased as well. Therefore, the less activated (and less stiff) the muscle, the better its ability to attenuate the acceleration produced by the impact force (Derrick, Hamill & Caldwell, 1998; Flynn, Holmes & Andrews, 2004; Holmes & Andrews, 2006). Holmes & Andrews (2006) have suggested that the soft tissues in the leg may be able to reduce the acceleration values in locations proximal to the initial impulse. For example, the proximal structures of the tibia experienced a decrease in acceleration in proportion to a decrease in muscle activation levels.

Given the proportional relationship between applied force and the resulting acceleration, many leg studies that have focused on force attenuation have used surface mounted accelerometers to more accurately measure the transient force characteristics of impact forces acting on the limb. An advantage of using accelerometers is that they do not require assumptions regarding initial velocities or starting positions (Wakeling & Nigg, 2001). Nevertheless, this method has yet to be implemented in the upper extremities of living subjects.

To effectively produce impacts to the upper extremity, Chiu and Robinovitch (1998) and DeGoede et al. (2002) have designed two structurally different fall simulations capable of describing the outcomes of an impact to the upper extremity. Chiu & Robinovitch (1998) used a free fall scenario that represents a safe, relatively natural fall configuration. However, unwanted, variable movements may occur given the inherent attempt to protect oneself against such a fall. DeGoede et al. (2002) attempted to overcome this source of variability by repositioning the subjects in an upright seated position. Although safe and effective, this method unnaturally impacted the hand, as subjects were asked to manually arrest a moving mass. The method of hand first impacts proposed in this study, combines the benefits of these methods by creating a safe and effective impact mechanism, while minimizing unwanted motion and creating a more natural impacting configuration. Similar human pendulum methods have proven successful for impacting the lower extremities (LaFortune & Lake, 1995; Flynn, Holmes & Andrews, 2004; & Holmes & Andrews, 2006). To date, this method has not been used in impact research with the upper extremity.

A clear understanding of how the upper extremity responds to impact is essential for preventing injury and for designing strategies and products that are focused on lowering the forces experienced during such events. Although the efficacy of wrist guards has been extensively researched and debated, these studies are limited in their sole reliance on cadaver specimens that lack the active and passive tissue properties of live subjects. Furthermore, upper extremity impact research has focused on the impact outcomes at the impacting surface only. However, as many leg studies have indicated, the impact shock wave has the ability to travel to more proximal anatomical locations with the soft tissues capable of affecting the shock wave outcome, a concept not yet studied in the upper extremity.

Statement of Research Questions.

Therefore, the purpose of this thesis was to answer the following research questions:

1. What are the acceleration response characteristics of the wrist and elbow following a hand-first impact resulting from a simulated forward fall?
2. What are the effects of manipulating forearm muscle activation level and elbow angle on wrist and elbow response following these types of impacts?
3. Is the implementation of a wrist guard effective in altering the response of the wrist and elbow following hand first impacts?

Hypotheses

Wrist Guards

H_0 = There will be no difference in the PA, AS, TPA, between the un-guarded and guarded conditions at the wrist or the elbow measured in the axial and off-axis directions.

H_0 = There will be no difference in the IRF or kinetic energy bewteen the un-guarded and guarded conditions.

Elbow Angle

H_0 = There will be no difference in the PA, AS, TPA, between the straight arm and natural arm conditions at the wrist or the elbow measured in the axial and off-axis directions.

H_0 = There will be no difference in the IRF or kinetic energy between the straight arm and natural arm conditions.

Muscle Activation

H_0 = There will be no difference in the PA, AS, TPA, between the four levels of muscle activation (baseline, 24 %, 36 % & 48 %) at the wrist or the elbow, measured in the axial and off-axis directions.

H_0 = There will be no difference in the IRF or kinetic energy between the four levels of muscle activation (baseline, 24 %, 36 %, & 48 %).

Chapter II
Review of Literature

General Background

Slips and trips are a continuing societal problem that can often lead to falls. Nevitt & Cummings (1993) found that the greatest frequencies of falls are those that occur in the forward direction. It is estimated that occupational injuries related to slips trips and falls can result in direct costs that exceed US $ 6 billion annually (Chang, 2002). Often the term slips, trips and falls (STF) is used to describe a variety of occurrences, however it is important to understand that these are all separate events. Bentley & Haslam (1998, 2001) suggest that an individual may slip (because of poor friction between the foot and supporting surface), trip (interaction with an external obstacle) and that the result is a fall, either on the level (occurring from the same height as gait), or from a height (from an elevation to a lower surface). Bentley et al. (2005) found that 65% of the fallers they studied cited their inability to move freely in their environment as the main reason for falling. The impact, which results from a fall, is considered the injury causing mechanism, and therefore has received considerable attention. Much of the research concerning forward falls, and the injuries to the upper extremity which occur as a result, has focused its attention on three "at risk" populations: the general work force, the elderly, and in-line skaters.

General Work Force

WSIB (2004) reported that 22.8 % of lost time injury claims were a result of a fall with 12.8 % occurring "on the level". Furthermore, the upper extremity was the affected area 22.7% of the time. Layne and Pollack (2004), in their investigations of non-institutionalized civilian workers reported that 15 % of injuries were a result of STF.

Over 66 % of the falls occurred on the level, and 72.1 % resulted in a complete fall, impacting the floor. Concerning affected body part, 18 % of hospitalized injuries occurred to the upper extremity in the form of a non-specified fracture or sprain. Although only 7 % percent of the injuries reported by Bentley & Haslam (1998) occurred to the upper extremity, 19 % of these resulted in greater than three days absence from work. As further illustrated by Courtney & Webster (1999) and Chang (2002), fall-related upper extremity injuries seem to be the most disabling as seen by the increase in days away from work.

In-line Skaters

Given the unstable nature of the activity, in-line skaters are also at a high risk of upper extremity injury. The distribution of injuries with respect to age is somewhat different from that of the population in industry. Here the distribution is bimodal, with the first peak of injuries occurring between the ages of 16 and 20 years, and a second peak occurring between the ages of 46-55 (Jaffe, Dijkers & Zamentis, 1997). Ellis et al. (1995), Houshian & Anderson (2000), and Houshian, Mehdi & Larssen (2001) found that falls associated with in-line skating resulted in a higher frequency of more serious injuries, with upwards of 50 % resulting in a fracture. Furthermore, Jaffe et al. (1997) reported that 64% of the upper extremity injuries required some form of orthopaedic surgery. With respect to injury location, Scheiber et al. (1996), in a nation wide study, found that 32% of the injuries occurred to the wrist with 25 % resulting in a fracture. The elbow accounted for 9 % of the injuries with 6 % of these ending in a fracture. Houshian, et al. (2001) found that at the elbow, 50 % of the injuries occurred to the supracondylar humerus, 10% to lateral epicondyles, and another 10% to the olecranon.

Elderly

Age-related deficiencies have been identified by numerous authors as risk factors for falls and fall-related injuries (Nevitt & Cummings, 1993; Cummings & Nevitt, 1994; O'Neil et al., 1994; Tinneti & Williams, 1998; Pavol et al., 1999; Palvanen et al., 2000; Layne & Pollock, 2004; Cherry et al., 2005). The frequency of wrist fractures in older adults increases between the ages of 50 and 65 years, and then appears to plateau (Nevitt & Cummins, 1993). Although wrist fractures remain the most common upper extremity injury in older adults, the frequency of injuries to the proximal humerus has been found to be increasing. In a study by Palvanen et al. (2000), 90% of proximal humerus fractures, 91% of wrist fractures, and 90% of elbow fractures were a result of a fall from standing height.

The effects of forward falls onto the outstretched arm are a constant risk throughout the lifespan of an individual. The physical consequences of such events have profound direct and indirect costs associated with them, both monetary and in terms of quality of life (Donald & Bulbitt 1999). As a result, it is apparent that research must move forward to reduce the outcome effects of a forward fall in all populations.

Upper Extremity Anatomy

The upper extremity is a relatively complex system of joints and segments consisting of hard tissue, in the form of bone, and soft tissues in the form of muscles, tendons, ligaments, and subcutaneous adipose. The shoulder, the elbow and the wrist are the largest joints of the upper extremity and are the joints most often involved in upper extremity injury (Figure 1) (All figures in this section are adapted from Tortora, 2005).

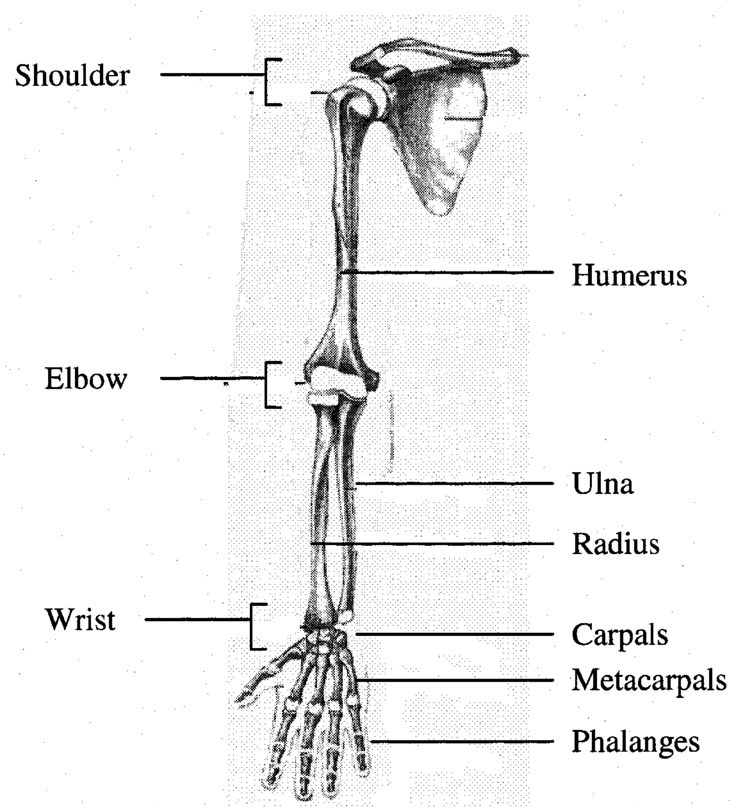


Figure 1. General description of the upper extremity.

The Shoulder

The shoulder, or gleno-humeral joint, is formed by the articulation between the head of the humerus and the glenoid fossa of the scapula. The rotator cuff muscles, consisting of the teres minor, infraspinatus, supraspinatus and subscapularis provide the majority of support to the joint. Other muscles, such as the pectoralis major, latissimus dorsi, teres major and deltoid are also involved in significant movement of the humerus about the shoulder (Figure 2 & Figure 3).

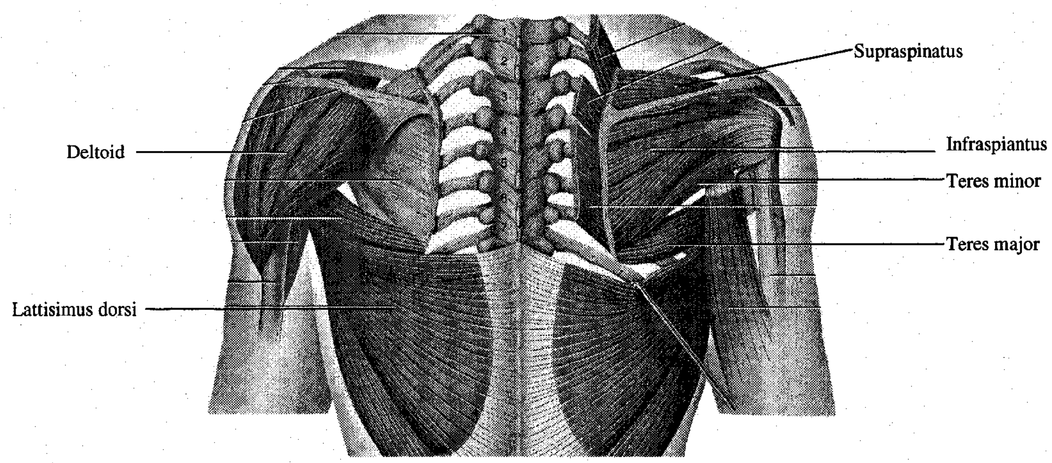


Figure 2. Posterior muscles that move the humerus about the shoulder.

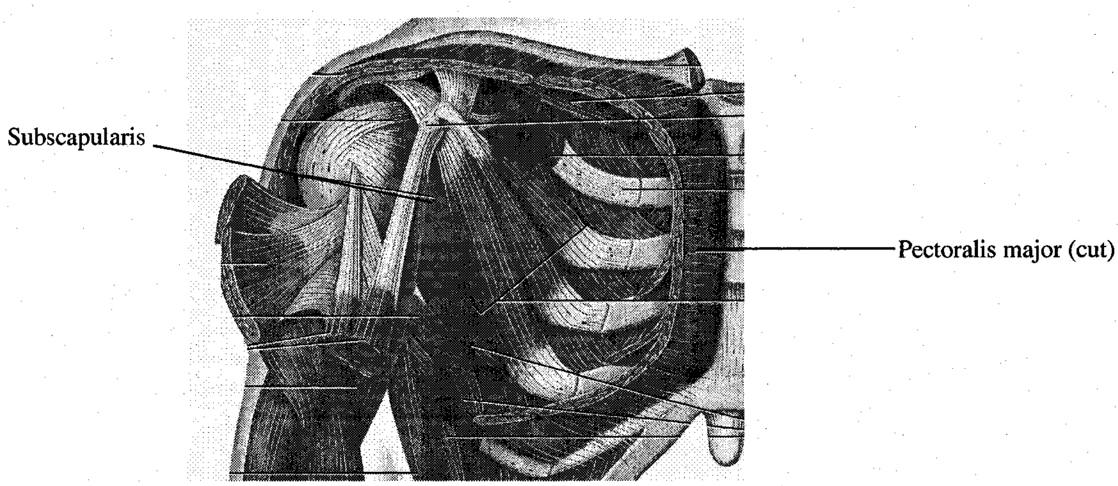
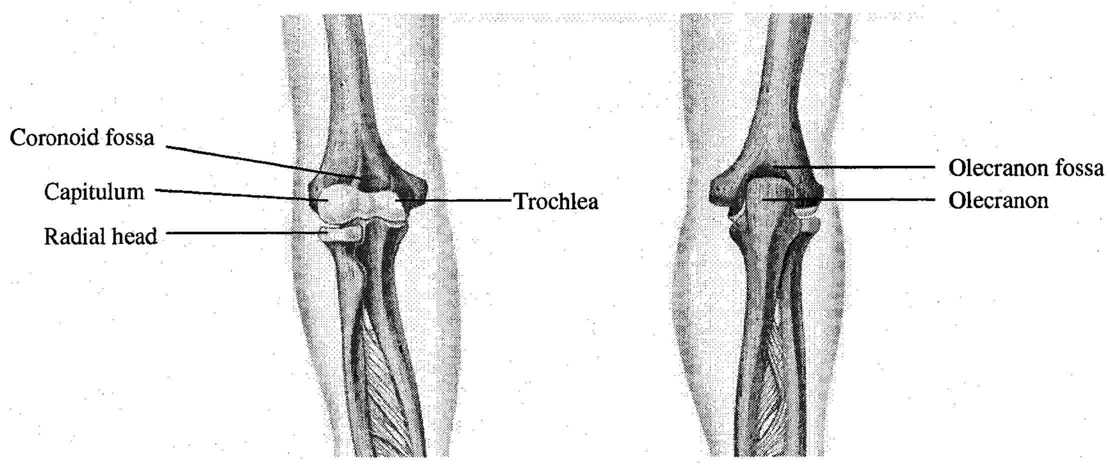


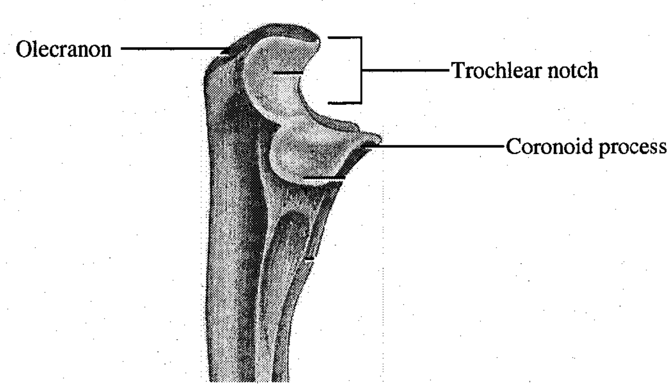
Figure 3. Anterior muscles that move the humerus about the shoulder.

The Elbow

The main sagittal plane movements of flexion and extension of the forearm are made possible by the interaction between the trochlea of the distal humerus and the trochlear notch of the proximal ulna. Furthermore, the forearm can be fully extended, as the olecranon process of the ulna is accepted by the olecranon fossa, a depression on the posterior, distal aspect of the humerus (Figure 4).



(a)



(b)

Figure 4. (a) bony structures that compose the elbow. (b) head of the ulna, showing the main articulating structures.

Lastly, the radial head glides against the capitulum throughout the range of motion of the elbow. The elbow gains support medially from the ulnar collateral ligament, laterally from the radial collateral ligament and anteriorly from the articular capsule and annular ligament.

The Wrist

The radius and ulna extend distally from the elbow where the radius articulates with the carpals to form the wrist joint (Figure 5). The ulna is positioned medially while the radius is positioned on the lateral aspect of the forearm. Both a proximal and distal articulation occur between the radius and ulna (radio-ulnar joints), allowing the radius to pivot and cross over the ulna when the forearm is pronated and supinated.

An interosseous membrane confines the intermediate space between the radius and ulna and is a common site of attachment for the deep musculature of the forearm. Protruding distally from the radius and ulna are the styloid processes. The radial styloid is a common site of fracture often referred to as a Colles fracture (McKinley & O'Loughlin, 2006). The wrist joint itself is formed by the interaction of two of the carpals (scaphoid and lunate) with the distal end of the radius. There are two additional proximal carpals, triquetrum and pisiform, while the distal portion of the wrist consists of four carpals (trapezium, trapezoid, capitate and hamate) that are responsible for forming the articulations with the metacarpals of the hand (Figure 5).

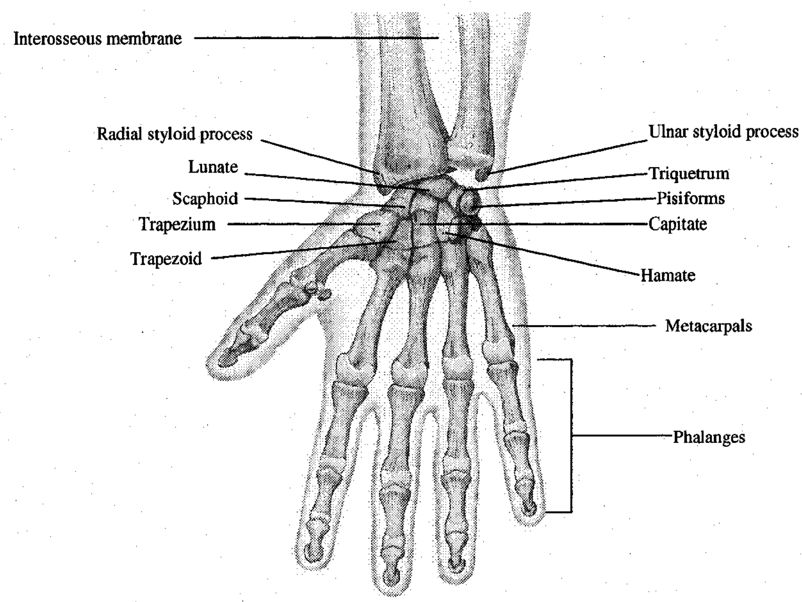


Figure 5. Bones of the wrist including the articulation with the radius.

Wrist Musculature

There are approximately 50 muscles located in the upper extremity, 18 of which are located in the posterior and anterior forearm and are responsible for the movement of the wrist and fingers. The bulk of the muscle is situated in the proximal two-thirds of the forearm, with these muscles giving way to long tendons that extend distally to their respective insertions. Four of these muscles, flexor carpi ulnaris, flexor carpi radialis, extensor carpi ulnaris and extensor carpi radialis, are of significant importance as they are the main contributors to flexion and extension of the wrist and help to form the large superficial mass of muscle in the proximal forearm (Figure 6). Anteriorly, the two flexor muscles share a common origin on the medial epicondyle of the humerus. Flexor carpi ulnaris travels distally where it inserts onto pisiform, hamate, and the fifth metacarpal. Flexor carpi radialis migrates diagonally to the inferior medial (radial) aspect of the forearm where it inserts onto the second and third metacarpals. Moving medio-laterally over the carpals is the flexor retinaculum, a structure responsible for ensuring that the tendons remain in close contact with the bony structures preventing the tendons from "bowing" when the muscles are activated.

Located on the posterior aspect of the forearm are the extensor muscles: extensor carpi radialis and extensor carpi ulnaris. Extensor carpi radialis diverges into two divisions known as brevis and longus which continue inferiorly and laterally to insert on the second and third metacarpals, respectively. The extensor carpi ulnaris however, moves from its origin on the lateral epicondyle of the humerus, across to the posterior forearm and inserts onto the fifth metacarpal. Similar to the flexor muscles group, the tendons of the posterior extensor muscles are held securely in place distally by the extensor retinaculum.

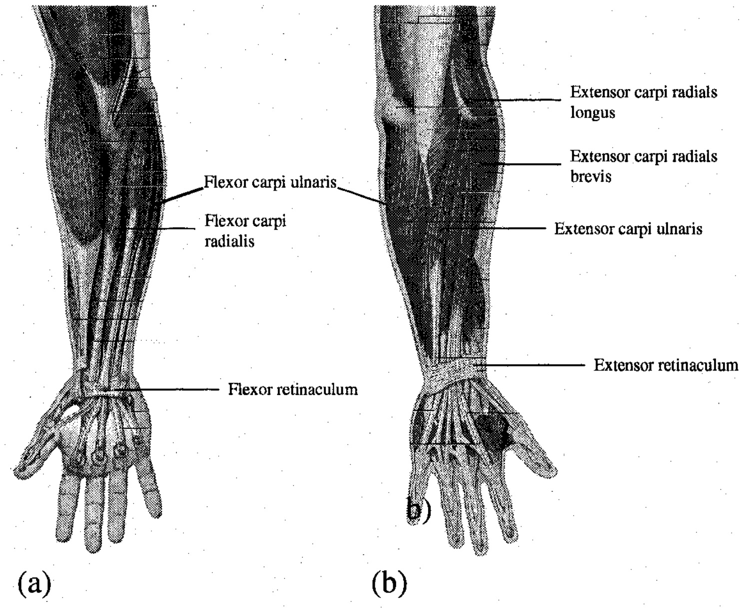


Figure 6. Anterior (a) and posterior (b) muscles that move the hand about the wrist.

Situated intermediate and anterior to the distal aspect of the carpals and the base of the first and fifth metacarpals, is a significant mass of palmar soft tissue composed of four muscles extending from the flexor retinaculum and tendons of the flexor muscles

(Figure 7). Abductor and flexor polices brevis are responsible for abducting and flexing the distal phalanx of the thumb, respectively. The abductor digiti minimi causes abduction of the fifth phalanx while flexor digiti minimi brevis is responsible for flexion of the fifth phalanx. Upon impact to the upper extremity during a forward fall, it is this mass of muscle that commonly comes into contact with the impacting surface first (Nikolic, Hancevic, Hudec, & Banovic, 1975).

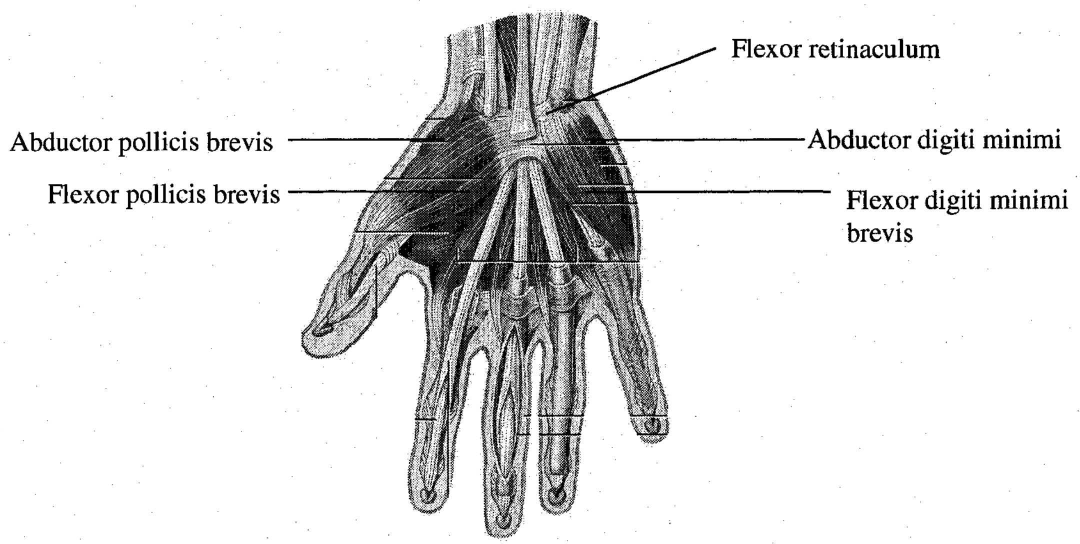


Figure 7. Anterior muscles of the hand, forming the palmar soft tissue.

Biomechanics of Forward Falls

Understanding the biomechanics of falls and fall-related outcomes serves a twofold purpose: it allows for the precise understanding of the mechanisms and structures involved in arresting a forward fall and, it may help lead to the development of interventions and protective mechanisms aimed at reducing the risk of injury related to forward falls. More information must be gathered concerning the external biomechanical demands which are applied during an impact so that they can be compared to the biological capacity of the involved structures (bone and soft tissue). This will allow researchers to predict whether an injury will occur, measured in terms of an injury risk ratio. This measure takes into account the demand placed upon the tissue and the tissue's ultimate capacity. When demand exceeds capacity (represented by a ratio greater than one), it is more likely that an injury will result (DeGoede, Ashton-Miller & Schultz, 2003).

Fall Phases

Biomechanically, the forward fall can be described by the following three, distinct phases: i) a perturbation phase, ii) the descent phase and iii) the impact phase. Although we are mostly concerned with the impact phase, because the risk of injury to the upper extremity is greatest during this phase, the first two phases ultimately dictate how the impact will occur.

Perturbation Phase

In the first phase, an individual must have their balance perturbed, causing a loss of stability and an initiation of descent. Pavol et al. (2001) determined the underlying causes for being unable to regain balance prior to a fall as being: the failure to reestablish an adequate base of support, inability to rotate the leg forward fast enough, excessive forward rotation of the Head Arm Torso (HAT) complex, and walking speed just prior to initiation of the trip. Pai & Patton (1997) also suggest that the ability to maintain balance is a function of both the velocity and position of the centre of mass (CM) of the trunk relative to the base of support defined by the feet. Using a pendulum model, they determined upper velocity-position boundaries that are necessary to avoid a forward fall. For example, this model predicted upper boundaries of 0.80 m/s while the CM is positioned over the heel and 0 m/s if the CM is over the toe. That is, if the velocity states exceed these limits given their respective positions, the CM will continue forward resulting in a forward fall. These conditions are initiated by either a trip or slip, as were defined in earlier sections.

Descent Phase

Once balance has been lost, the faller begins the descent phase, which is characterized by the upper extremities reaction in preparation for the landing and impact. Hsiao & Robinovitch (1998) and Kim & Ashton-Miller (2003) have reported extensively on this phase and the movements which comprise it. During this phase, reported as being approximately 0.40 – 1.0 s in duration (Hsiao & Robinovitch 1998; Kim & Ashton-Miller, 2003; Tan, Eng, Robinovitch & Warnick, 2006), and aptly named the available reaction time (ART), individuals select a desired response to the fall in an attempt to arrest their generated momentum. Kim & Ashton-Miller (2003) manipulated falling distance (0.4-1.0 m), thereby increasing the ART, and reported the subsequent upper extremity kinematics. The upper extremity pre-impact kinematics followed a similar pattern, for all fall distances. These movements include the arm being raised in front of the body via elbow extension and shoulder flexion, such that a "stiff arm" posture was attained. They suggested however, that at greater fall heights, where there is an obvious increase in the ART, excessive shoulder flexion, and elbow extension, and less wrist extension occur at touchdown. Although this "stiff arm" posture has been associated with greater impact forces (Chou, et al., 2001; DeGoede & Ashton-Miller, 2002; DeGoede, Ashton-Miller, Schultz & Alexander 2002; DeGoede & Ashton-Miller, 2003; Lo, McCabe, DeGoede, Okuizumi, & Ashton-Miller. 2003), it is speculated that it is adopted to prevent injury to the head, pelvis and hip regions. Hsiao & Robinovitch (1998) support the findings of Kim & Ashton-Miller (2003) as they also found that an outstretched hand posture was adopted during the descent phase. The movements generated during the descent phase are crucial in ascertaining how the individual will impact the ground.

Impact Phase

During the impact phase, the faller makes contact with the ground and arrests the momentum (and therefore energy) of the body. The postures adopted during the descent phase govern how the hands will impact the ground and the resulting ground reaction forces (GRF). Originally, Hsiao and Robinovitch (1998) found that ground contact occurs either simultaneously to the hands and knees or to the hands first and shortly after to the knees (approximately 50 ms after hand contact), suggesting that contact to the knees has no bearing on the hand GRF.

Impact Characteristics

Upon contacting the ground, the upper extremity serves to arrest the falling mass (the human body) and absorb the energy associated with it. In its simplest form, the energy associated with the fall is a function of the initial height which can be illustrated through the conservation of energy. Post-contact, the upper extremity behaves as a mass spring damper system, to arrest and dissipate the energy of the body (Chiu & Robinovitch, 1998). The upper extremity impact force can be characterized by two distinct transient force peaks (Figure 8). The first, "high-frequency" peak occurs immediately after hand contact and represents the initial impact of the upper extremity ($Fmax_1$). The second, "lower-frequency" peak, occurs a short time after the $Fmax_1$ and represents the upper extremities attempt to slow or "break" the momentum of the body ($Fmax_2$) (Figure 8).

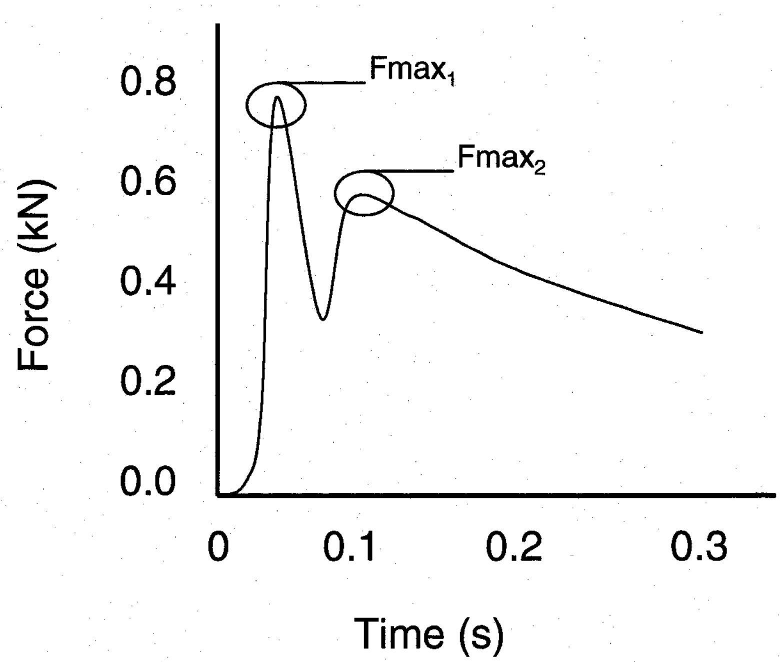


Figure 8. Force trace of an upper extremity impact, adapted from Chiu & Robinovitch (1998).

Studying only straight-armed falls, from three different, relatively low heights (1, 3 and 5 cm), Chiu & Robinovitch (1998) found that the initial force peak ranged from 0.4 to 0.8 kN and was proportional to impact velocities between 0.2 m/s – 0.8 m/s. The first peak was found to occur approximately 20 ms after impact while the second peak was delayed until approximately 110 ms post impact (90 ms after $Fmax_1$). The two peaks were also found to be influenced by different variables. For instance, descent height was found to increase $Fmax_1$ by an average of 390 N from 1-5 cm, but only influenced $Fmax_2$ by 36 N between the three heights. Furthermore, body mass was also found to have significant effects on the force outcomes. $Fmax_1$ was only affected by body mass at heights of 1 and 3 cm, whereas $Fmax_2$ was positively related to body mass at all three descent heights ranging from 0.6-0.8 times body weight. $Fmax_1$ is representative of the transfer of potential energy at the hand-ground interface. Therefore, it is a function of both mass and height. With mass remaining constant, descent height is the main contributor to the initial force peak. With the majority of energy being dissipated throughout the initial contact, $Fmax_2$ is representative of arresting the mass of the body, now supported by the upper extremity, and with very little change in height occurring.

<u>Elbow Angle Effects</u>

Chou et al. (2001) and DeGoede et al. (2002) suggest that elbow flexion during impact may help to reduce the GRF at the wrist, thus providing individuals with a preventative strategy to reduce injury. Subjects were instructed to either maintain an extended elbow, similar to Chiu & Robinovitch (1998), or to impact the surface with the elbow flexed. Flexion conditions differed between these studies; Chou et al. (2001) had only a flexion and extension condition (flexion was subject to the individuals' preference), whereas DeGoede et al. (2002) incorporated three elbow angles (170°, 150° and 130°) at three different heights, simulated by adjusting the velocity of the pendulum mass (1.8 m/s, 2.3 m/s, 3.0 m/s). DeGoede et al. (2002) were able to provide a relationship between elbow angle and peak force. Impact force increased by values of 1.4 N/deg, 1.9 N/deg and 2.3 N/deg at 1.8m/s, 2.3m/s and 3.0 m/s, respectively. Conversely, Chou et al. (2001) were unsuccessful in showing that elbow posture was capable of reducing the overall GRF. However, they were able to show a decrease in some of the more specific joint forces of the upper extremity, specifically elbow antero-posterior and medio-lateral shear components. Despite their inability to show a decrease in force between flexion and extension conditions, Chou et al. (2001) provide a relationship between the occurrence of elbow flexion and the time to peak force. Two different time-to-peak force scenarios were observed; individuals either flexed the elbow immediately upon impact, in which case maximal impact was delayed until $Fmax_2$ or, elbow flexion occurred after impact, where loading occurred in the same pattern as an extended elbow posture. An initial thought regarding these results was that the greater time to peak force may provide time to react to the impact and further their strategy to resist an injury. Upon further investigation, it was noted that the entire time of impact is only a few ms, hardly enough time to consciously react, and form a preventative strategy.

Furthering the analyses of elbow angle on impact force, DeGoede & Ashton-Miller (2003) created four upper extremity impact scenarios: i) a natural fall where individuals where asked to arrest the fall with no instructions, ii) a non-simultaneous condition, where subjects were asked to retract the right hand just prior to impact, allowing the left hand to arrest the fall in isolation, iii) a "stiff arm" approach, similar to

Chiu & Robinovitch (1998), with the additional verbal queue that subjects were to arrest the fall so that they kept their head as far off the ground as possible, and iv) a "minimum impact" trial, where individuals were instructed to arrest the fall and to minimize the peak force obtained. Results showed that maximum forces were created during the stiff-arm trials, a result that again supports the findings of Chiu & Robinovitch (1998). Individuals were able to lower the peak force by approximately 19%, from 1021 N to 831 N, for the stiff-arm and natural landing techniques, respectively. Furthermore, the peak force was reduced by 40 % overall when the individuals used a "minimizing" strategy. The same patterns of force decline were presented for $Fmax_2$. These data provide further parametric insight into how the upper extremity reacts during impact. The elbow angle at the time of initial impact for all four strategies differed only by a maximum of 12°, ranging from 162° for the non-simultaneous condition to 174° for the "stiff-arm" condition. Average elbow angle for the "natural" strategy was reported to be 168°. However, at the time of the second peak, "natural" elbow angle decreased to approximately 155°, and finally to approximately 120° at the end of impact. The second important finding was that there were no significant differences between the non-simultaneous and natural hand condition. The trailing hand contacted the force plate approximately 40 ms after the first hand made contact. The difference in time was only found to delay the impact characteristics of the trailing hand and did not change the magnitude of the forces recorded. This suggests that similar forces will occur at the hand whether the hands come into contact with the impact surface simultaneously, or independently. Further analyses on the timing and force characteristics of these events without a verbal queue, will provide insight to the mechanics of a forward fall.

Impact Methods

Chiu & Robinovitch (1998) implemented a "free fall" strategy that kept the knees on the ground (creating a 30° angle between the ground and the thigh), and the fall was arrested unilaterally with one outstretched arm (15° from the vertical). The falls occurred from 1, 3, or 5 cm above the ground. Similarly, Chou et al. (2001) and Tsai, Chou, Chou, & Lin (2003) adopted this method in their own attempt to analyse the underlying consequences of a forward fall from a height of 5 cm.

DeGoede & Ashton Miller (2002) and Lo et al. (2003) used a method that took the individual off their knees and initiated a fall in a semi-upright position. Individuals were lowered and asked to flex forward so the shoulders were approximately 1m above the ground. Unlike Chiu & Robinovitch (1998), a number of upper extremity postures were used, thus creating a variety of shoulder and elbow angle combinations. However, all falls were carried out such that only the hands contacted the ground (i.e. neither knee or torso contact occurred).

Kim & Ashton-Miller (2003) and Tan et al. (2006) both used the same initial starting position, however with contradicting outcomes. Kim & Ashton-Miller (2003) varied fall distance by moving the individual backwards at intervals of 20 cm, over a range from 40-100 cm. In this case, the impact surface was located in front of the faller approximately 1.2 m off the ground, allowing subjects to fall into the platform. In comparison, Tan et al. (2006) let the individual fall from standing height to the ground, with an initial forward lean of approximately 15° to the vertical. Individuals were instructed to contact the ground with flexed knees before commissioning the upper extremity to arrest the fall.

Implementing another approach, DeGoede et al. (2002) used a swinging pendulum mass to initiate contact with the upper extremity while the subjects were seated in an upright position. The individuals were instructed to arrest the mass using a variety of shoulder-elbow posture combinations, while simulating three different fall heights. Fall height was manipulated by altering the velocity of the pendulum (1.8 m/s, 2.3 m/s and 3.0 m/s). The seated position provided adequate support to the lumbar region of the spine while allowing adequate movement of the upper extremity.

Although different in structure, these methods have proven successful in simulating the events that occur during a forward fall, and more importantly, those events which take place after impacting the upper extremity. However, there are distinct pros and cons to each of the methods previously mentioned. The "free fall" method creates a natural falling configuration where the individual moves toward the impacting surface. It is instinctive to protect oneself from falling and becoming seriously injured, and therefore, this method may create unwanted movement in the impacting segments (arm, forearm, scapula), and may vary between individuals. DeGoede & Ashton-Miller, (2002) moved the subject to an upright sitting posture that allowed them to more accurately control segment movements. The impact surface in this method however, is traveling toward the subject; a situation different from a naturally occurring fall. The human pendulum method proposed in this thesis, combines the benefits of the previous impact mechanisms. While the impacts are safe, and maximally controlled, they represent a more traditional approach with the subject traveling towards the impact surface.

Impact Modeling

Due to the risks and complications associated with in-vivo impacts, at distances representative of falling from standing height, impact models have been developed to simulate these falls. These models allow researchers to estimate the forces experienced at greater heights and make conclusions regarding the risk of injury. Two models of significant importance were developed by Chiu & Robinovitch (1998) and DeGoede et al. (2002), respectively.

Chiu & Robinovitch (1998) developed a linear, two mass, spring damper impact model (Figure 9). Both the mass of the torso and the upper extremity are represented in this model as m_2 (mass of the torso) and m_1 (mass of the upper extremity). Although this model takes into consideration the relative damping and stiffness of the shoulder/torso complex (k_2, b_2) and the palmar soft tissue (k_1, b_1) it assumes that the impact occurs to a straight arm void of any elbow flexion.

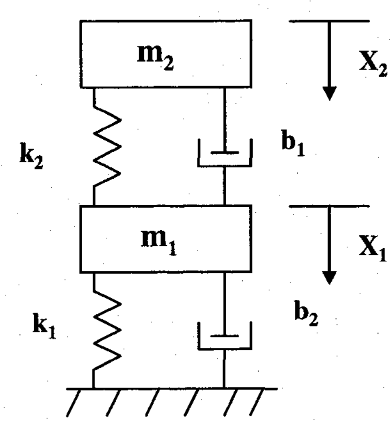


Figure 9. Linear two mass model used to calculate hand impact force (Chiu & Robinovitch, 1998).

Upon impact, the system as a whole (m_1 and m_2) moves through a distance of x_1, determined by the damping and stiffness properties and representing the initial impact force ($Fmax_1$). The first mass is eventually brought to rest, however, and the stiffness and damping of the shoulder now attempt to arrest the mass of the torso, thus theoretically producing a shoulder impact force.

The forward dynamic model, developed by DeGoede et al. (2002), represents a non-linear mass spring damper system which incorporates elbow angle at the instant of impact (Figure 10). This system was initially designed to determine heel contact in running, but has recently been adapted to the upper extremity. In this case the impact is initiated by a swinging lumped mass (M) contacting the hands (M_h). Similar to Chiu & Robinovitch (1998) the effects of the palmar soft tissue are accounted for with stiffness and damping coefficients (b_h & k_h). The elbow joint is represented by the union of two rigid segments (representing the arm and forearm) joined by a torsional spring (k_e) that represents the restoring torque that takes place after impact occurs. The shoulder is represented by a mass-less slider (S), the movement of which is under the control of the stiffness and damping properties (b_s & k_s).

These models are both capable of describing the events that occur during an impact to the upper extremity, and predicting the potential forces that occur at the hand. However, these models are designed to calculate the force effects at the hand primarily, and are not capable of describing the impact effects at more proximal locations such as the elbow. In addition, calculating the hand forces is a daunting task, requiring first the determination of the stiffness and damping coefficients of the involved tissues. Finally, these models assume that the upper extremity is a rigid linked system and does not take into account the passive properties of the soft tissues in the limb.

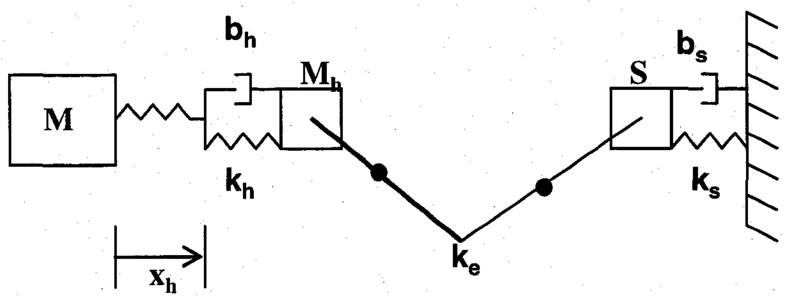


Figure 10. Non-linear hand impact model incorporating initial elbow angle (DeGoede et al., 2002).

Measuring Bone Response

Measurement

Accelerometers are capable of measuring the shock produced by an impulse, traveling through skeletal structures (Lafortune, Hening, & Valiant, 1995). The optimal method of collecting data regarding the acceleration of the bony structures is through bone mounted accelerometers (BMA) that are attached directly to the underlying bones via intracortical pins. However, when live human subjects are involved, the use of BMA are not always an option and researchers must rely on less invasive techniques such as surface mounted accelerometers (SMA). Kim, Voloshin, Johnson, & Simkin (1993), and Lafortune et al. (1995) measured the accuracy of using SMA by comparing them directly to the outputs obtained from BMA. Their concern was that the soft tissue underlying the SMA would distort the signal and provide inaccurate measurements. When proper protocol is followed however, SMA is a suitable surrogate for BMA for measuring the acceleration of the underlying bony structures. Kim et al. (1993) reported the most accurate results when lightweight accelerometers were pre-loaded, (i.e. strapped down) within the subjects' pain threshold, onto the underlying bone. The acceleration values collected by an accelerometer are proportional to the applied force. Therefore, a properly employed surface mounted accelerometer is sensitive to the movement of the underlying bony tissue, and can provide insight into the effect of a more distally applied impact force.

There are three components of segment acceleration-time profiles that are of particular interest when determining impact effects (Figure 11). These dependent variables are of interest as they provide insight into the forces occurring as a result of an impact and the rate at which the anatomical structures are being loaded at. The peak acceleration (PA) represents the maximum value obtained over a particular time (measured in Gs; equal to the number times the gravitational acceleration of 9.81 m/s^2). Depending on the accelerometer in use, it is possible to measure force effects in all three orthogonal planes. The other dependent variables of acceleration are the time to peak acceleration (TPA) and the acceleration slope (AS). The AS measures the rate of change of acceleration with respect to time and has been previously measured as the tangent between 30 % and 70 % of the PA, which ultimately represents the rate of loading of the impacted segment (Flynn, Holmes & Andrews, 2004; Holmes & Andrews, 2006).

Upper Extremity Acceleration Studies

Surface mounted accelerometers have been extensively used to study lower extremity impact situations. To date, upper extremity impact studies have relied on the derivation of kinematics and force plate data to make inferences regarding proximal joint outcomes. However, these methods require one to make assumptions regarding the initial velocity and position parameters (Wakeling & Nigg, 2001). Furthermore, as Light, McLellan & Klenerman (1980) suggest, the true effect of the impact at more proximal locations is difficult to comprehend simply by relying on what occurs at the initial location of impact.

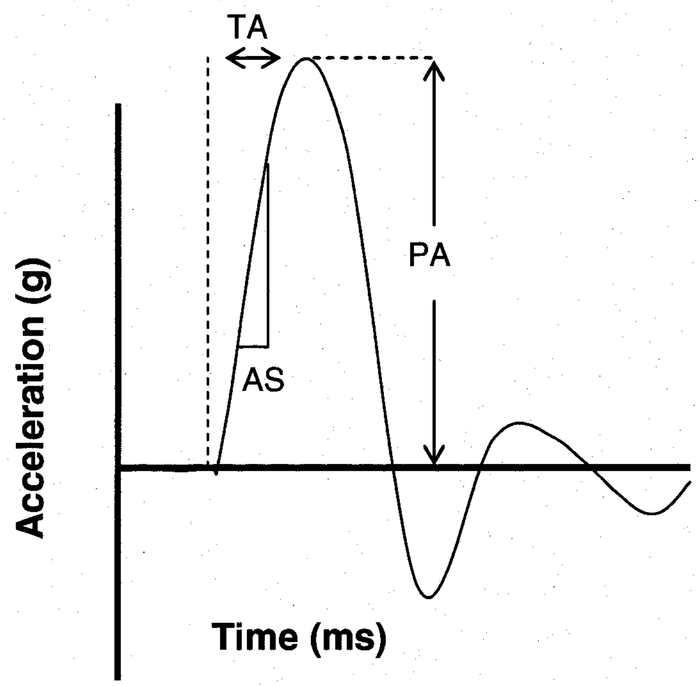


Figure 11. An actual acceleration waveform measured at the tibial tuberosity along the longitudinal axis of the shank post impact. Copied with permission from Holmes & Andrews (2006).

Only two studies have attempted to quantify the impact effects in the upper extremity through the use of accelerometers. Hwang & Kim (2004) positioned the forearm in an inverted pendulum formation with the elbow affixed to the surface and the hand capable of moving in the sagittal plane. Acceleration of the dorsal aspect of the hand was recorded, as the hand impacted the ground. Although this movement does not correctly simulate a forward fall, force and velocity variables were consistent with those that occur during impact, with hand acceleration values averaging 43.1 Gs. One should take caution in interpreting these results as they represent the acceleration at the back of the hand and not the wrist, or other segments, associated with impact arrest.

While studying the impact effects of airbag deployment on the outstretched arm, Atkinson et al. (2001) were successful in reporting proximal and distal forearm, and, proximal and distal humerus accelerations. Peak wrist resultant accelerations were measured at 456 Gs and impact velocity was approximately 4 m/s. The highest values recorded occurred while the forearm was pronated and the wrist was extended a posture that is consistent with the moment of impact during a fall. Acceleration of the proximal forearm was shown to be negligible. However, unlike a forward fall the force impulse lasted only 19 ms.

The lack of information regarding more proximal structures of the upper extremity suggests that more needs to be known regarding the impact response at the levels of the elbow and shoulder, for living human subjects. Atkinson et al. (2001) also suggest that acceleration data can provide a quick and accurate tool that can be used to assess the risks of upper extremity impacts.

Soft Tissue Properties

Impact forces travel as shock waves through tissues such as bone, fat, muscle, tendons and ligaments, moving away from the initial area of impact (Chu, Yazdani-Ardakani, Grandisar & Askew, 1986; Light et al., 1980; Wosk & Voloshin, 1981). These studies shed light on the behaviour of the shock wave, concluding that certain structures in the body act as attenuators and are capable of dissipating the energy of the shock, such that the force experienced at more proximal structures is less than the initial impact. The soft tissues, collectively regarded as wobbling mass (Gruber, Ruder, Denoth, &

Schneider, 1998) are responsible for the passive attenuation of the shock wave. However, the body is also capable of actively attenuating the shock wave by changing the angles of joints (Gerristen, van den Bogert & Nigg, 1995; DeGoede et al., 2002). Authors agree that muscle is the most important force attenuating structure given its ability to passively deform, thereby absorbing the impact energy, while simultaneously possessing the ability to actively adjust the amount of shock attenuation through eccentric contractions about the joints (Elftman 1939; Derrick, Hamill & Caldwell, 1998; Pain & Challis 2001). An example of the passive attenuating capabilities of muscle was presented by Nikolic, Hancevic, Hudec, & Banovic (1975) who found that the soft tissue of the palm was capable of attenuating the force prior to reaching the distal radius, similar to the effects of the fat pad found in the foot (Pain & Challis, 2001).

Muscle Activation

Changes in muscle activation, potentially leading to a change in muscular stiffness, plays a key role in the attenuation of force and the dissipation of energy within and between body segments. When a muscle's level of activation is increased, it is representative of the increase in the tension generated in the muscle tissue itself. This effect is created by the interaction of a greater number of cross bridge units (Mijailovich, Fredberg & Butler, 1996; Lee et al., 2006) It is this tension increase that is responsible for the subsequent changes in structural stiffness (Winter, 1991; Nigg & Liu, 1999;). Laquinta, Licata, & Soechting, (1982) and Osu & Gomi (1999), studying the upper extremity, were both able to provide evidence that as the musculature of this segment was activating to greater levels, the stiffness about the elbow joint proportionally increased as well. Laquinta, Licata, & Soechting (1982) went even further to state that both the elastic and viscous properties of the muscle were increased. The relationship between muscle activation and stiffness has also been applied to the trunk, with similar findings as the upper extremity studies (Lee, et al., 2006).

It is thought that the body has an ability to adjust the level of muscular activation in a protective response to a perturbation. This strategy is often referred to as muscle tuning (Wakeling, Nigg, & Rozitis, 2002; Boyer & Nigg, 2004). Wakeling, Nigg & Rozitis (2002) generally regard the stiffness of the tissue as being proportional to the level of force that is transmitted through it. Specifically, it is changes in the damping and frequency characteristics of the muscle, that leads to the muscle's ability to alter the characteristics of the impact shock wave (Wakeling & Nigg, 2001; Wakeling, Nigg, & Rozitis 2002). Therefore, the relationship between the aforementioned variables can be stated as: an increased stiffness achieved through the increased activation of a specific muscle group leads to an increase in the force that is transmitted to adjacent structures in the body, that is, less of the force is dissipated in the soft structures (mainly muscle).

Muscle stiffness changes have been altered with various methods. For example, Nigg & Liu (1999) simulated changes in muscle stiffness, while impacting the leg during running. Flynn, Holmes & Andrews (2004) used fatigue to alter the muscle's ability to generate tension, while Holmes & Andrews (2006) experimentally altered muscle activation levels (15%, 30% 45% & 60% of maximal voluntary exertion). Both of these latter studies analysed the effects of an impact in the leg. These studies found that as muscle activation levels decreased, therefore decreasing the stiffness of the segment, the peak accelerations displayed at the tibial tuberosity were found to decrease. In addition, Flynn, Holmes & Andrews (2004) noted that the acceleration slope was also decreased as activation levels were decreased as a result of fatigue, suggesting the tibia is loaded at a slower rate as a result. Understanding the force attenuating properties of the wobbling mass is an important concept as it allows us to more accurately assess the effects of impacting the segment (Gruber et al., 1998; Pain & Challis 2001).

The upper extremities contain force attenuating tissues similar to those found in the shank, suggesting that shock waves may also be attenuated as they travel more proximally and may vary with differing levels of muscle activation (DeGoede et al., 2002). To date, the effect of various muscle activation levels on the forearm response to impact and wave travel through the upper extremity remains unknown.

Biological Capacity

Wrist Capacity

The ability of a tissue to withstand loading without experiencing failure is an inherent function of the tissue's structure. With respect to bony tissue, the most commonly injured tissue during impact, there are a number of properties that lend to its ultimate strength. For instance, Augat, Reeb & Claes (1996), and Meyers, Hecker, Rooks, Hipp & Hayes (1993) found that specific geometric variables such as thickness and area of the cortical shell, and the bone mineral density of the bone all contributed to the ultimate strength of the bone. Furthermore, as suggested by Greenwald, Janes, Swanson, & McDonald (1998) the viscoelastic properties of bone, which are highly dependent upon the rate of loading, such that, the tissue becomes much stiffer if it is loaded at a faster rate, also have an affect on the failure load of bony tissue.

Many researchers have mechanically loaded cadaver specimens to determine the maximal load strength of human forearms, specifically the distal radius (Frykman, 1967; Horseman, Currey, & Phil, 1983; Meyers et al., 1991; Meyers et al., 1993; Augat, Reeb & Claes, 1996; Giacobetti, Sharkey, Bos-Giacobetti, Hume & Taras, 1997; Greenwald et al., 1998;). With the exception of Greenwald and colleagues, who dynamically loaded cadaver forearms, all others used quasi-dynamic loading to impact the forearm specimen.

A summary of loading rates and failure loads are found in Table 1, as adapted and modified from DeGoede et al. (2003). The average failure load was approximately 2619 N, with a range of 1580 N – 3773 N. The large range in variability can be attributed to the different loading rates that were used by different researchers. Frykman (1967) proposes a classification system for distal radial fractures based on the location of the fracture with respect to the articulating surfaces (radio-carpal and distal radio-ulnar) which are classified as either intra-articular or extra-articular. This classification system also takes into consideration whether there is a fracture of the distal ulna. All together there are eight classifications for distal radial fractures as presented by Frykman (1967).

Table 1. Summary of failure loads of the distal radius representing the capacity of bone tissue.

Author(year)	Loading Rate	Mean Failure Load (N)
Men		
Frykman (1967)	16 N/s	2770
Meyers et al., (1991)	25 mm/s	3740
Meyers et al., (1993)	25 mm/s	2370
Augat et al., (1996)	1 mm/s	3773
Women		
Frykman (1967)	16 N/s	1917
Horseman et al., (1983)	0.8 mm/s	3600
Meyers et al., (1991)	25 mm/s	3180
Meyers et al., (1993)	25 mm/s	1580
Augat et al., (1996)	1 mm/s	2008
Unknown Gender		
Spadardo et al., (1994)	0.4 N/s	1640
Spadardo et al., (1994)	0.4 mm/s	2410
Giacobetti et al., (1997)	25 mm/s	2245
Greenwald et al., (1998)	Dynamic impact	2821
Troy & Grabiner (2007)	Unreported	2752 (Axial) 1448 (Off-axis)

Elbow Capacity

Traditionally, the focus of injury risk prevention and assessment has been the distal radius, and to a lesser extent the distal ulna. However, as the epidemiological evidence suggests, the elbow joint (olecranon of the ulna, head of the radius and the condylar region of the humerus) may also be susceptible to injury as a result of falling on the outstretched arm (Houshian, Mehdi, & Larsen, 2001). The mechanism of injury to this region is a more complex issue than at the wrist because of the interactions of the three bones that comprise the elbow joint. Many have identified various disorders that may occur as a result of axial impact forces. Injuries to the elbow are generally referred to as fracture-dislocation injuries and are classified with respect to the location of the fracture (Wake, Hashizume, Nishida, Inoue, & Nagayama, 2004). Supracondylar fractures occur distally in the humerus in the area superior to the epicondyles. This type of injury is classified with respect to the degree of displacement as well as the degree of cortical fracture (Housain, Mehdi, & Larsen, 2001). Coronoid fractures are classified with respect to the size of the fragment broken off from the coronoid process (Ablove, Moy, Howard, Peimer & S'Doia, 2006; Doornberg & Ring, 2006). Finally, another structure susceptible to injury is the olecranon, which most commonly experiences a translocation-fracture. These fractures are classified based on the degree of comminution of the trochlear notch and the proximal ulnar metaphyseal area (Ring, Jupiter, Sanders, Mast, & Simpson, 1997).

Wake, Hashizume, Nishida, Inoue & Nagayama (2004) attempted to quantify the forces required to produce fractures in the structures of the elbow. Cadaver specimens of intact elbows were loaded at 0.16 mm/s, and failure was found to occur between 300 N –

700 N, nearly a third of the force required to produce an injury at the wrist. Of all the specimens tested, fracture-dislocations were found in the humeral shaft (13%), supracondylar or condyle (30%) and the radial or ulna shaft (28%). In addition, 50% of the specimens suffered a fracture-dislocation specific to the humeroulnar joint in conjunction with one of the injuries listed above.

Understanding the relationship between demand and capacity at all levels of the forearm is crucial in attempt to prevent upper extremity impact injuries. Furthermore, these data, both from capacity studies and external demand studies, have established physiological (failure loads) and biomechanical (velocity, position and force) parameters for future research. Physiological parameters ensure that subjects are impacted safely at loads below those that are required to cause a serious injury such as a fracture. Biomechanical parameters allow us to compare our results to past research and make certain that accurate, reliable data are being collected.

Injury Prevention Strategies

To date, the common goal of prevention strategies has been to reduce the GRF at the moment of hand contact or to dissipate the energy through a surrogate material other than biological tissue. Three methods of force reduction, wrist guards, fall interventions and compliant surfacing, have all been proposed to lessen the effects of forward fall impacts.

Wrist Guards

The purpose of wrist guards is to transfer force and energy away from the wrist, most often with a rigid, plastic, volar splint. Secondly, the wrist guard functions to resist extension of the wrist; a posture most often adopted to arrest a fall and associated with a high incidence of Colle's fracture (Frykman, 1967; Meyers et al., 1993). Giacobetti et al., (1997), Hwang et al., (2006), Staebler et al., (1999), Greenwald et al., (1998), Kim et al., (2006) have all researched the mechanics of wrist guards including force diversion and energy absorption. Hwang & Kim (2004) and Kim et al. (2006) analysed a variety of materials which are most effective in reducing the impact force to the wrist. However, there appears to be conflicting opinions on the issue of the efficacy of wrist guards. For instance, Giacobetti et al. (1997) found no statistical difference in fracture force among those specimens with and without guards, and found similar fracture patterns between the groups. Cheng, Rajarantnam, Raskin, Hu, & Axelrod (1995) reported on four cases of individuals who had sustained a forearm fracture while using wrist guards. It was suggested that the rigid nature of the device, which can often be worn up to the mid-forearm, transfers the energy to this location, leading to a form of forearm fracture.

Furthermore, Hwang et al. (2006) found that wrist guards were only effective in preventing fractures in lower impact/energy situations and only during backwards falls.

Scheiber et al. (1996), Staebler et al. (1999), Greenwald et al. (1998), and Hwang & Kim (2004) have all provided evidence supporting the notion that implementing wrist guards are effective in reducing hand impact force. It was found in these studies that the use of a wrist guard could decrease the force of a fall by 39 to 61%. Other findings suggested that the impact velocity (Greenwald et al., 1998) and the energy (Hwang & Kim, 2004) could also be lowered while using a wrist guard as protection.

These studies are limited by their reliance on cadaver specimens, and self-reported data. Further data are required on live subjects, before conclusions can be made regarding the efficacy of wrist guards as a preventive measure.

Fall Training

From the data collected by DeGoede & Ashton-Miller (2002) with regards to elbow angle effects, Lo et al. (2003) were able to study the effectiveness of a fall intervention. The aim of this intervention was to train individuals to fall in a protective manner. Similar training has been used in the area of martial arts, however this was the first attempt to protect the individual while falling only in a forward direction. Individuals who received the ten-minute intervention were able to adopt proper falling techniques and lower the impact force by approximately 18%, immediately prior to the training session. However, after a three-month retest, there were no significant differences in impact forces between the intervention and non-intervention groups.

Fall Surface Properties

Similar to the materials of a wrist guard, the impacting surface may also have an effect on the impact force. An effective surface must be able to dissipate energy and transfer force away from the impacting segment, therefore limiting the forces most likely to cause injury. Several epidemiological studies have examined the relationship between severity of injury and the type of surface the impact occurs against (Nevitt & Cummings, 1993; Mowat, Wang, Pickett & Brison, 1998). Robinovitch & Chiu (1998) experimentally studied the effects of surface stiffness on the GRF applied to the wrist. Using different thicknesses of foam padding to vary the stiffness of the impacting surface, subjects were dropped using similar methodology as in Chiu & Robinovitch (1998). Stiffness levels ranged from 252 kN/m (most stiff) to 53 kN/m (least stiff). The force experienced at $Fmax_1$ was decreased by approximately 34% when comparing the absence of protective surfacing (force plate without any padding) to a surface with stiffness of 252 kN/m. However, further decreasing the stiffness had no significant effect on the wrist force. Furthermore, it was found that not only are compliant surfaces capable of attenuating forces, but they are also effective in delaying the time to peak force. Although protective surfacing seems an enticing solution, there remains a delicate balance between protecting the upper extremity and maintaining a surface that does not disrupt normal gait (i.e a surface that is too compliant).

The common factor among these devices is their ability to attenuate and delay the applied force. However, all have focused their attention on decreasing the force applied to the wrist, and have ignored the effects that may occur in more proximal locations.

Summary

In summary, the above review of literature clearly shows the need for further exploration into the mechanical workings and response of the upper extremity to impact. To date, valuable data have been recorded with regards to upper extremity impacts, specifically to the wrist only. Research into protective mechanisms to lower the effects of the resultant forces has also received considerable attention. However, as the data suggest, there are other, more proximal structures of the upper extremity that are affected by impacts, and there are clearly other biological components involved in the attenuation of the resulting forces. Research concerning lower extremity impacts has clearly shown that both active and passive tissue properties have an effect on the impact shock wave and its ability to travel to more proximal locations. Therefore, it seems relevant to assume that the upper extremity may react in a similar manner and that it is necessary to quantify the characteristic responses of the forearm following impacts to the hand.

Chapter III
Methodology

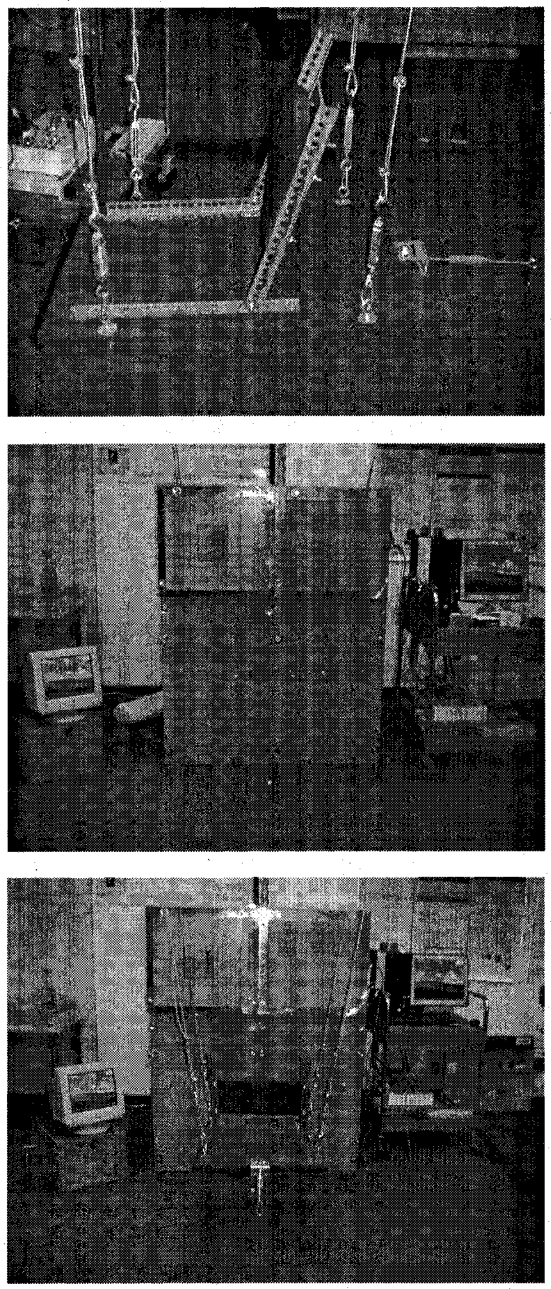
Subjects

A total of 28 subjects (15 male and 13 female) with mean (SD) age 22.3 (2.31) years, mass 72.4 (13.2) kg, and height 1.70 (0.09) m were recruited from the University of Windsor, Human Kinetics student population. The mean (SD) body mass index (BMI) of the study sample was 24.9 (3.71) kg/m^2 . All subjects were free of upper extremity injuries at the time of testing. Individuals were asked to read and sign an informed consent prior to participation. The methods and procedures were approved by the Research Ethics Board of the University of Windsor.

Materials

Pendulum Apparatus

A seated human pendulum was used to simulate the flight phase and the impact phase of a forward fall (Figure 12a). The seat (77cm x 53 cm x 30 cm) was made from 6 mm (1/4") plywood, covered in 3 mm (1/8") foam padding. 2.5 cm (1") perforated angle iron made up the seat frame which in turn was bolted to the seat. The entire apparatus was suspended from the ceiling by four aircraft cables of equal length, attached to each corner of the seat such that the seat was situated approximately 0.6 m above the ground. The individuals faced towards the impact surface so that the angle between the trunk and the horizontal was maintained at approximately 110°-120°; a posture typical of a forward fall (Chiu & Robinovitch, 1996; DeGoede & Ashton-Miller, 2002; Tan et al., 2006). A backrest was positioned at the level of the lumbar spine (30 cm from the seat) to provide support, but also allow subjects to control their upper extremity segments during the impact phase.



(a)

(b)

(c)

Figure 12. Experimental setup including (a) the pendulum apparatus (b) the impact apparatus design with two force plates, and (c) the pendulum and impact apparatus complete with data collection computer and display computer.

The subjects were secured by two 5 cm straps; one was placed around the hips, and one was secured around the thighs and under the seat to support the legs in a flexed position. This configuration ensured that the trunk was adequately supported while placing the lower extremities in a comfortable position. This design allowed for the control of arm postures, which may otherwise alter the force values. The seat pan width and depth were designed to accommodate up to the 90^{th} % ile male, based on thigh length and intertrochanteric width, respectively (Pheasant, 1996).

<u>Impact Apparatus</u>

Two force plates were mounted to their own 2 cm thick steel plate (46 cm x 51 cm). The force plates and underlying steel plates (57 cm x 60 cm) were attached to a steel impact frame (153 cm x 122 cm x 127 cm), built from 4 cm square tubing and rigidly attached to the floor and wall (Figure 12b). The bottom of the force plates were located 83 cm from the floor. Two pieces of 6.54 cm x 6.54 cm steel tubing spanned the width of the impact apparatus extending the force plates away from the impact apparatus, allowing the knees to travel under the force plates and avoid contact with the impact apparatus. Foam padding was also placed around the bottom of the impact apparatus to protect the knees should they impact the apparatus following hand impact.

Instrumentation

<u>Force Plates</u>

Force data were collected in the Fz direction (i.e. in the direction of progression) using two vertically mounted force plates (AMTI-OR6-6-1000, A-Tech Instruments Ltd, Scarborough ON, Canada) that were bolted to 2 cm thick steel plates (56 cm x 61 cm).

These data were used in conjunction with the velocity values (see below) to ensure that impacts were occurring within normal physiological ranges.

Velocity/Position

Position and velocity data were used to verify that the swing distance was comparable with drop heights discussed in the literature. A distance that produces a force of 41% BW and a velocity of 1 m/s was used. The velocity and position of the pendulum was collected using a linear velocity/displacement transducer (Celesco DV301, Don Mills, ON, Canada), which was attached to the trailing edge of the pendulum.

Electromyography

Kendall bipolar disposable Ag/Ag-Cl surface electrodes (23 mm x 33 mm) were placed over the muscle bellies of the main wrist flexor and wrist extensor muscle groups (2 cm inter-electrode distance) of the right forearm in the direction of their lines of action (Figure 13). The activation levels of the wrist flexor muscle group (e.g. flexor carpi ulnaris) located anteriorly, and the wrist extensor muscle group (e.g. extensor carpi ulnaris) on the posterior aspect of the forearm, were collected. The electrodes were placed as follows: flexor carpi ulnaris – two fingerbreadths from the ulnar border on the proximal third of the forearm, extensor carpi ulnaris – just lateral to the ulnar border of the mid-forearm, and extensor carpi radialis – two fingerbreadths distal to the lateral epicondyle (Mogk and Kier, 2003). All EMG data were full wave rectified, and filtered with a 2^{nd} order, dual pass Butterworth filter (cut-off of 1.5 Hz), and normalized to the subject's Maximal Voluntary Exertion (MVE) level (see Procedures section).

Figure 13. Placement of the instrumentation on the right arm of the subject.

Electro-Goniometer

Elbow angles were measured by a twin axis electro-goniometer (Biometrics SG110 Biometrics Ltd., Gwent, UK). With the forearm pronated, the proximal head of the electro-goniometer was placed just proximal to the lateral epicondyle, and the distal head was placed over the lateral forearm. The electro-goniometer was held in place with two-sided tape. A computer screen in view of the subjects during testing displayed the target elbow angle and the subjects' actual elbow angle (see Procedures section).

Accelerometers

Accelerations of the distal radius (representing the wrist) and the proximal ulna (representing the elbow) were measured to determine the transient force effects with respect to the specified conditions. Two surface mounted micro-machined tri-axial accelerometers (MMA1213D and MMA3201D, Freescale Semiconductor, Inc, Ottawa

ON, Canada) with a range of +/- 50 G and +/- 40 G, respectively, were used. The accelerometer signals were filtered close to the source with a 4^{th} order Bessel switched capacitor filter. The accelerometers were firmly attached to the skin using double sided tape and a normal pre-load of approximately 45 N, which was applied using an elastic Velcro™ strap. The distal accelerometer was placed on the posterior surface of the distal forearm just medial to the radial styloid process. The proximal accelerometer was placed over the olecranon process of the proximal ulna. The shape of the forearm did not allow for the accelerometers to be perfectly aligned. Only two axes were analyzed at each location: the sensitive axis, parallel to the long axis of the forearm, and an off-axis, acting at a right angle to the proximal and distal radius. At the wrist, the sensitive axis accelerations were always positive representing a transient, proximally oriented shock wave. The off-axis accelerations were also always positive, representing a force that is transmitted through the hand and towards the dorsal surface of the hand, at a right angle to the carpals. The sensitive axis acceleration response at the elbow however, could be both positive and negative. The positive output represents the same transient effect as at the wrist; however, a negative response is representative of a force oriented distally along the forearm. The off-axis response also had the potential to be either positive or negative dependent on the direction of the applied force. A positive response represents a force travelling at a right angle to the olecranon, oriented inferiorly. Conversely, a negative response is directed superiorly into the olecranon process (Figure 14). As such, acceleration data from each location were only compared along the same axis. The sensitive axes were visually aligned parallel with the longitudinal axis of the forearm.

Three dependent variables from the measured acceleration waveforms were analyzed. These include: peak acceleration (PA), time to peak acceleration (TA), and the acceleration slope (AS), measured between 70% and 30 % of the peak.

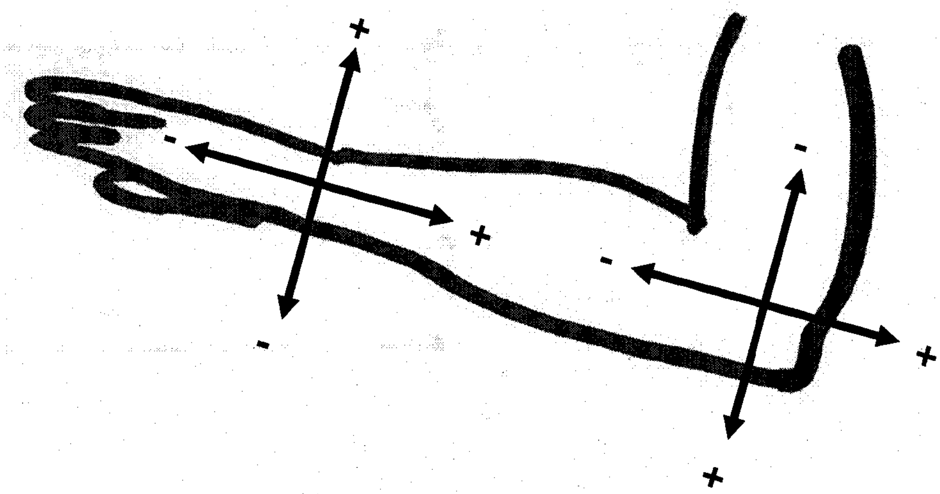


Figure 14. Schematic representation of the axes at the wrist and elbow. Included are the acceleration responses and the orientation of the force transients specific to each area.

Data Collection/Acquisition

Data were collected using a custom LabVIEW software program (National Instruments, Austin, TX, USA, version 7.1). Data collection was initiated manually by triggering the software to record acceleration, velocity, force, EMG and electro-goniometer waveforms simultaneously. All data were sampled at 4096 Hz and A/D converted with a 12-bit card.

Experimental Design

A 2X2X4 (elbow angle X wrist support X muscle activation) repeated measures design was utilized. All data were collected in one session with subjects being impacted three times per condition, giving 48 total impacts (Figure 15). This study analyzed the effects of three independent variables (muscle activation, elbow angle and wrist guards) on 12 dependent variables (PA axial and off-axis, AS axial and off-axis, TPA axial and off-axis, impact force, kinetic energy and FCU muscle activation).

Elbow Angle

Two elbow postures, "straight-armed" and "natural", were studied. These postures were maintained throughout the duration of the impact. The straight-arm condition was meant to simulate a "worse case scenario" fall, where the forearm is fully extended at the elbow and the shoulder angle remained at 90° to the torso. This was controlled for by the height and tilt of the pendulum relative to the impact surface. The second condition was meant to simulate a natural falling strategy, which positions the elbow at approximately 168° (this represents the angle between the arm and forearm segments as per DeGoede & Ashton-Miller, (2002)). In order to maintain this elbow angle and wrist posture the shoulder flexion angle was decreased by approximately 12°.

Muscle Activation

Processed EMG data from the extensor carpi ulnaris were used to determine the effects of four levels of muscle activation on elbow acceleration: baseline level (approximately 12%), 24%, 36%, and 48 % of MVE. These levels were displayed to subjects on a visible computer screen as they executed voluntary isometric contractions. Data collected from flexor carpi ulnaris were analysed for changes in muscle activity during the impact, which suggests antagonistic muscle contribution.

Wrist Guards

The ability of a wrist guard to attenuate potential injurious impact induced accelerations was studied. A Firefly sport line wrist guard (model number: 065627) was used for the portion of the study that required the impact hand to be protected. These guards were made from foam padding, with a rigid plastic plate that was positioned over the proximal palmar surface. The plate extends from the metacarpals up the forearm. The entire wrist guard was held in place with two Velcro™ straps: a 2 cm wide strap crossed the dorsal aspect of the metacarpals while a 5 cm strap was positioned just proximal to the radial styloid. This wrist guard design was consistent with the wrist guards that have been used in previous studies. Two sizes of wrist guards were made available (small and large) to ensure that the guards fit each subject comfortably.

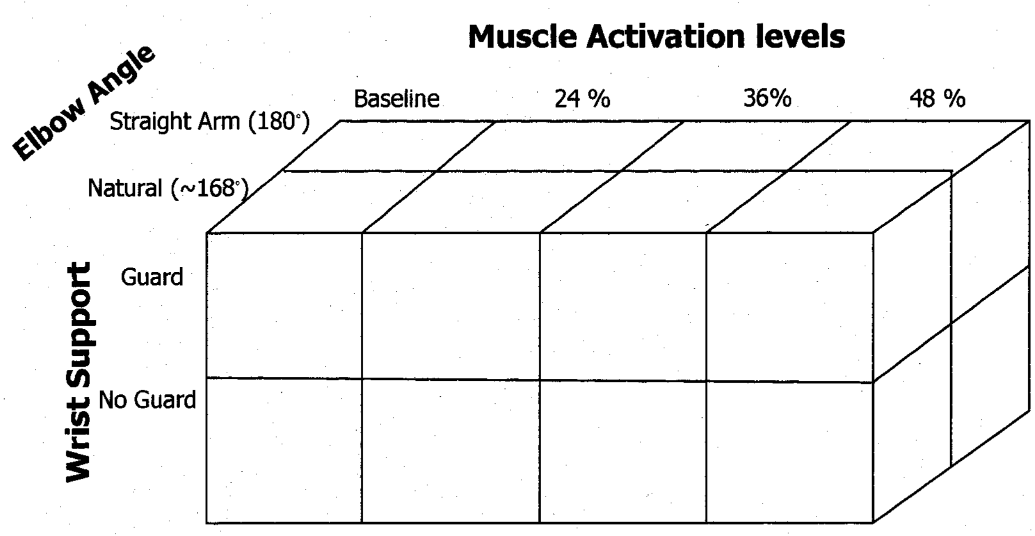


Figure 15. Schematic representation of the levels of each independent variable involved in the experimental design.

Procedures

Following explanation of the process, the subjects' right arms were fitted with the appropriate instrumentation (2 accelerometers, 2 channels of EMG, and an electro-goniometer). Maximal Voluntary Exertions (MVEs) were collected for all muscle groups prior to testing. MVEs were collected while the subjects were sitting with the forearm on the table flexed at 90° about the elbow and fully pronated. The hand was extended beyond the end of the table, allowing it to move through flexion and extension. Flexor MVEs were gathered by adding resistance to the palm of the hand and instructing the subject to flex the wrist. The hand was allowed to move through a limited range of motion, while being resisted, to ensure that the maximal possible activation was achieved (Potvin, 2001). Extensor MVEs were collected by asking the subjects to attempt to touch the back of the forearm with their middle finger. Limited resistance was applied to the back of the forearm to ensure that it was not lifted during the collection of MVEs. Three trials for each muscle group were collected and the peak EMG amplitude of all three trials was considered the individual's MVE. An adequate period of rest was given to the subject between each MVE trial to ensure that fatigue did not occur.

With the subjects seated in the pendulum, a "baseline" extensor EMG activation level was collected. Subjects were asked to place the palmar tissue of their hand lightly against the force plate, thus extending their wrist in the range of 30°- 40°, an angle consistently reported in the literature as the angle of the wrist that is most common at the moment of impact. The angle of the wrist was verified by the experimenter and the baseline EMG level was recorded as a percentage of MVE (mean baseline level was 15% across all subjects).

Prior to testing, a straight-arm angle was established for each subject. The straight-arm angle was collected in the same manner as the baseline activation level, and was recorded to ensure consistency within each subject during all trials.

Three to four trial impacts were carried out to ensure that the velocity was approximately 1 m/s and force values were between 40 % and 50 %. These parameters were consistent with previous research, with the subjects leaning forward just above the surface of a force plate approximately 0.10 m. The pendulum was manually retracted to the appropriate distance (approximately 0.4-0.5 m) which was marked for each subject so that consistent impact parameters were achieved.

Target elbow angles and target muscle activation levels were displayed to the subjects on a computer screen in their field of view. A randomly selected combination of these variables was determined prior to testing by the investigator. The pendulum was pulled back to the marker, and released following an auditory queue, which simultaneously started the data collection. In the no guard condition, impact occurred to the soft tissue area intermediate to the distal carpals and the base of the metacarpals, on the palmar aspect of the hand. The wrist was extended between 30° and 40°, and was regulated by the height of the pendulum relative to the force plate. Following completion of the elbow angle and muscular activation level treatments without a wrist guard, the subjects were fitted with the proper wrist guard and all elbow angle and muscle activation levels were repeated. The wrist guard trials followed the identical protocol as the unguarded trials. The wrist was able to extend against the guard while the wrist guard was in place throughout the impacts.

Statistics

Overall group averages for the three dependent variables, PA, TPA and AS, at each location and for both loading directions, were reported. A three way 2X2X4 (elbow angle X wrist support X muscle activation) repeated measures ANOVA was used to determine if significant main and interaction effects for the 12 dependent variables existed based on the experimental treatments. Significant main and interaction effects were considered for further analysis only if they accounted for at least 1% of the total variance (Keppel 1982). Tukey HSD testing was performed to post hoc significant effects. An analysis of the force plate data, including the maximal force attained ($Fmax_1$) and the time to $Fmax_1$, were carried out using t-tests to determine whether mean differences between impacts to the left and right hands were significant. Alpha was set at 0.05 for all comparisons.

Chapter IV
Results

Impact Force, Velocity and Kinetic Energy

Subjects impacted the force plates at an average velocity of 1.01 m/s and a force of 41 % body weight (BW) (331 N). A t-test revealed that the mean peak impact reaction force (IRF) of the right hand was significantly greater than the IRF of the left hand (41 % BW in the right vs. 32 % BW in the left) ($p<0.05$). Furthermore, in 82 % of the impacts, the right hand impacted approximately 0.4 ms prior to the left hand (Table 2).

The mean peak IRF for the right hand was significantly affected by the wrist guard, elbow angle and the different muscle activation conditions ($p<0.05$). IRF was decreased both when the wrist was guarded and when the elbow was flexed in the natural arm position. However, increasing muscle activation from baseline levels to 48 % resulted in an increase in the impact force by 29 % (Table 2)

The mean (SD) kinetic energy at impact was 41.7 J (0.38). Significant differences were found between the baseline level and 48% muscle activation ($p<0.05$). However, the overall difference in kinetic energy between these two levels was only 1.0 J on average (41.8 J vs. 42.8 J). A significant sex difference was also found; with males producing 14 % more impact energy than females (44.8 J vs. 38.6 J) ($p<0.05$) (Table 2).

Table 2. Mean (SD) impact forces, velocities and kinetic energies for all subjects (*p<0.05 across all levels). ¥ = force values for the right hand only.

		Sex		Wrist Guard		Elbow Angle		Muscle Activation Level				Hand	
	Overall	Male	Female	Guarded	Un-guarded	Straight	Natural	Baseline	24%	36%	48%	Left	Right
Force (%BW)	36 (10)	45* ¥ (3)	37 (3)	49* (2)	32 (1)	42* (2)	39 (2)	33* ¥ (1)	40 (2)	43 (2)	47 (2)	32* (7)	41 (10)
Kinetic Energy (J)	41.7 (1.6)	44.9 (1.6)	38.5 (1.7)	41.9 (1.2)	41.5 (1.2)	41.7 (1.2)	41.7 (1.2)	41.2* (1.2)	41.7 (1.2)	41.8 (1.2)	42.1 (1.2)	_	_
Velocity (m/s)	1.0 (0.4)	_	_	_	_	_	_	_	_	_	_	_	_

Electromyography

The four target Extensor Carpi Ulnaris (ECU) muscle activation levels were 12 % (baseline), 24 %, 36 % and 48 % of subjects' MVEs. Subjects were relatively accurate at reaching these targets as the mean (SD) corresponding muscle activation levels were 15 (0.7) %, 25 (0.2) %, 35 (0.2) % and 50 (0.9) %. No significant wrist guard or elbow angle effects were found, showing that subjects were capable of reaching and maintaining the desired level of muscle activation.

Electromyography was also monitored in the Flexor Carpi Ulnaris (FCU) to determine if there was any significant co-contraction. FCU EMG was not significantly affected by either the wrist guard or elbow angle conditions. However, FCU muscle activation increased significantly ($p<0.05$) across the four levels of ECU muscle activation, from 5 % at baseline, to 16 % at the highest level of muscle activation (Figure 16).

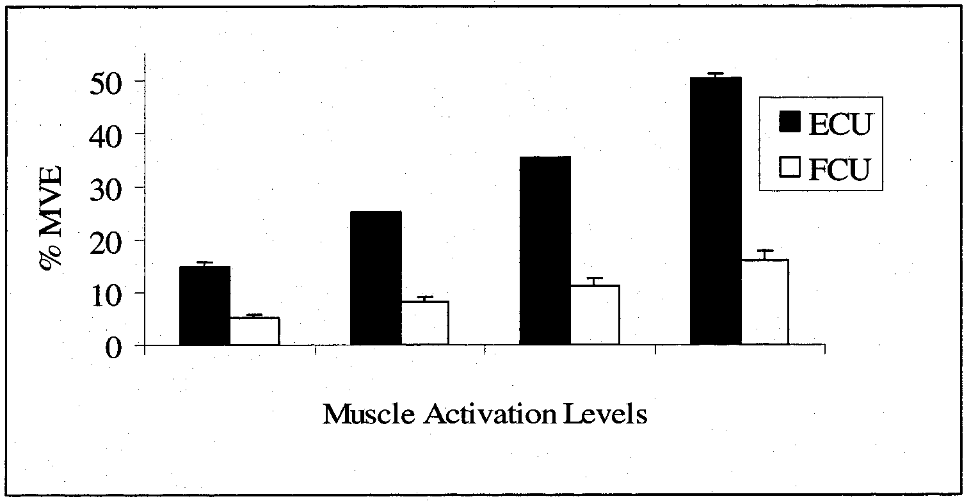


Figure 16. Comparison of Extensor Carpi Ulnaris (ECU) and Flexor Carpi Ulnaris FCU muscle activation levels. FCU significantly increased, as ECU was increased.

Acceleration

Wrist Response

Mean (SD) peak axial (PA_{axial}) and off-axis acceleration (PA_{off}) at the wrist were 11.4 g (0.5 g) and 15.2 g (1.1 g), respectively. Representative acceleration time histories are shown in Figure 17. The average rates of loading or acceleration slopes had magnitudes of 5454 g/s (321 g/s), directed axially along the forearm, and 613 g/s (72 g/s) directed off-axis (AS_{off}), or towards the dorsum of the hand. Time to peak acceleration occurred approximately 18 ms (0.1) axially (TPA_{axial}), and 24 ms (0.1) off-axis (TPA_{off}), post impact (Table 3).

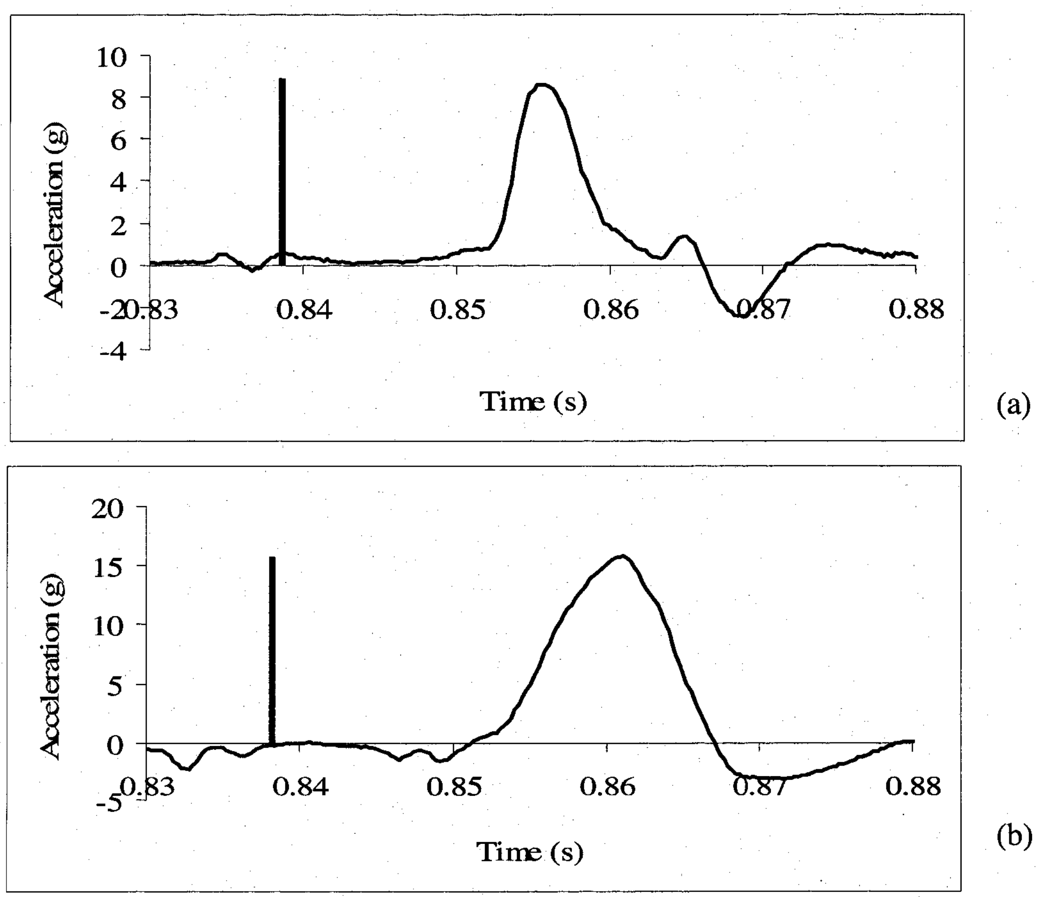


Figure 17. A representative axial (a) and off-axis (b) acceleration time trace measured at the wrist. The solid vertical line represents hand/force plate impact.

Wrist Guards

Wrist guard effects were dependent on the loading axis of the wrist. In general, there was no benefit of the wrist guard for protecting the wrist from axial loading. However, the wrist appeared to be significantly protected against loading in the off-axis direction by the guard that was tested in this study. Table 3 summarizes the mean data and highlights the main effects of the treatment conditions.

Peak Acceleration

No significant PA_{axial} differences occurred between the un-guarded and guarded conditions (Figure 18a) However, there were two two-way interactions with elbow angle and muscle activation that did significantly affect $PA_{axial.}$ The wrist guard by elbow angle ($p < 0.05$, $\omega^2 = 0.08$) interaction revealed that when impact occurred to the straight arm the PA_{axial} was increased by 1.0 g between the un-guarded and guarded hand. However, there was no difference between the un-guarded and guarded condition when impacts occurred to the natural arm (Figure 19).

Table 3. Mean (SD) acceleration response values at the wrist for all testing conditions. (*$p<0.05$, [a] $p<0.05$ with all other levels of muscle activation)

		Sex		Guard		Elbow Angle		Muscle Activation			
Acceleration Response	Overall	Female	Male	Un-guarded	Guarded	Straight	Natural	Baseline	24%	36%	48%
PA_{axial} (g)	11.8	11.9	11.7	11.6	12.0	11.5*	12.1	11.5	11.9	11.9	12.0
	(0.5)	(0.7)	(0.6)	(0.6)	(0.6)	(0.5)	(0.5)	(0.4)	(0.5)	(0.5)	(0.6)
PA_{off} (g)	15.2	15.4	15.0	20.1*	10.3	15.1	15.4	12.0[a]	15.0	16.2	17.6
	(1.1)	(1.5)	(1.4)	(1.8)	(0.5)	(1.0)	(1.2)	(0.9)	(1.1)	(1.2)	(1.2)
AS_{axial} (g/s)	5454	5292	5617	3193*	7716	5322	5587	4745	5301	5542	6230[a]
	(321)	(470)	(438)	(251)	(555)	(336)	(331)	(319)	(362)	(360)	(373)
AS_{off} (g/s)	4124	4698	3551	5642*	2607	3926*	4323	2643[a]	4049[a]	4613	5193
	(449)	(658)	(612)	(790)	(227)	(401)	(511)	(310)	(462)	(500)	(627)
TPA_{axial} (ms)	18	18	17	18	18	18	17	25[a]	15	15	16
	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(2)	(1)
TPA_{off} (ms)	24	24	24	22*	26	25	24	33[a]	23	21	21
	(1)	(1)	(2)	(1)	(1)	(1)	(1)	(1)	(1)	(3)	(1)

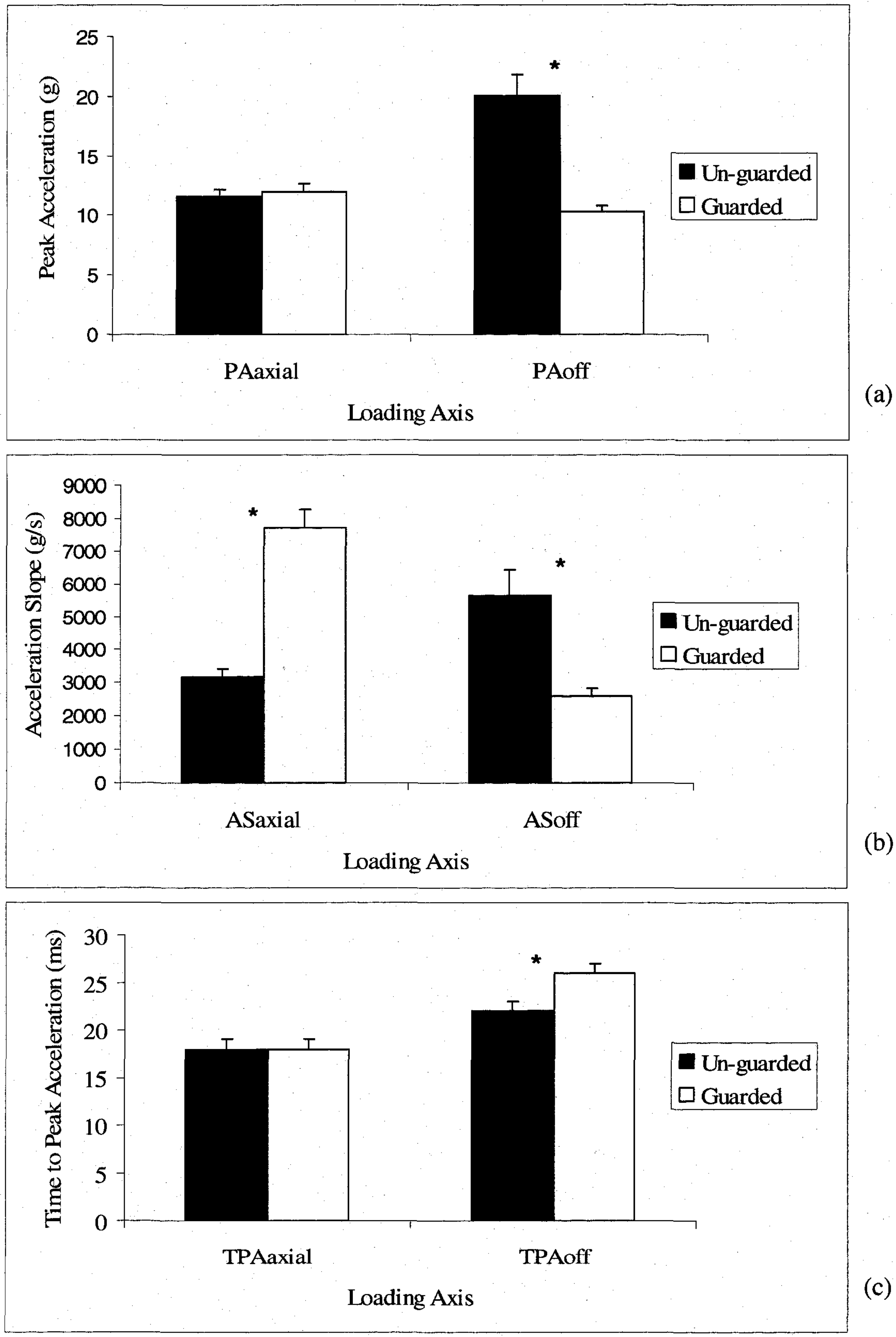


Figure 18. Mean peak axial (PA_{axial}) and off-axis (PA_{off}) accelerations (a), axial and off-axis acceleration slopes (AS_{axial} & AS_{off}) (b) and time to peak acceleration (TPA_{axial} & TPA_{off}) (c) at the wrist in the un-guarded and guarded conditions.

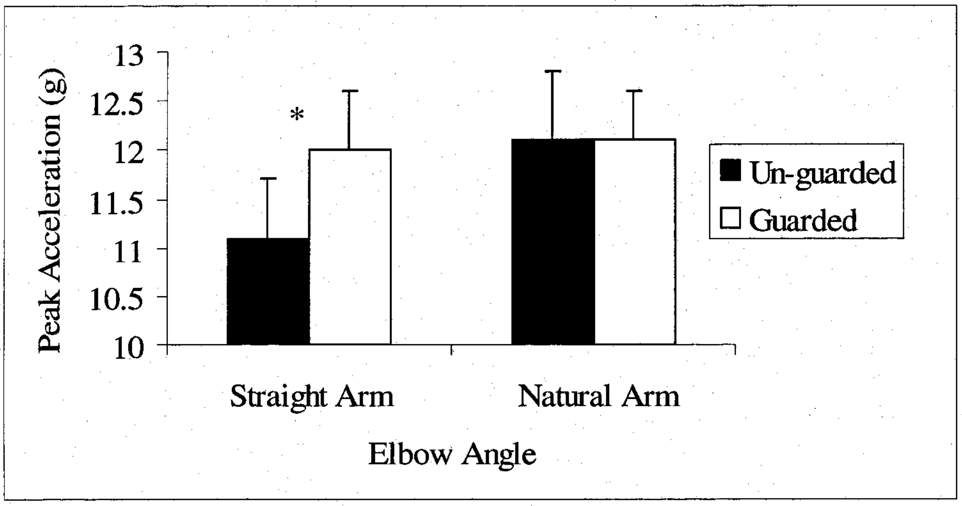


Figure 19. Wrist guard by elbow angle interaction effect at the wrist (*$p<0.05$).

The wrist guard by muscle activation interaction ($p<0.05$, $\omega^2 = 0.25$) showed a significantly greater PA_{axial} at baseline levels when a wrist guard was being used than when the wrist was un-guarded. Significant differences were not found at the 24 %, 36 % or 48 % muscle activation levels. Overall, PA_{axial} increased by 2.0 g from the baseline to the 48 % activation levels when the wrist was un-guarded and decreased by only 1.5 g when the wrist was guarded (Figure 20a).

There was a wrist guard main effect with PA_{off} almost 10 g or 49 % lower ($p<0.05$), compared to when no wrist guard was used (Figure 18a). This main effect was incorporated into a two-way interaction with muscle activation ($p<0.05$, $\omega^2 = 0.03$). The guarded wrist reduced PA_{off} by approximately 50 %, compared to the unguarded wrist, at each level of muscle activation (Figure 20b). At baseline, the un-guarded wrist experienced an acceleration of 15.1 g on average compared to only 8.8 g for the guarded wrist. At 48 % muscle activation the mean PA_{off} increased by 36 % (to 23.6 g), and 24 % (to 11.6 g) in the un-guarded and guarded wrist conditions, respectively.

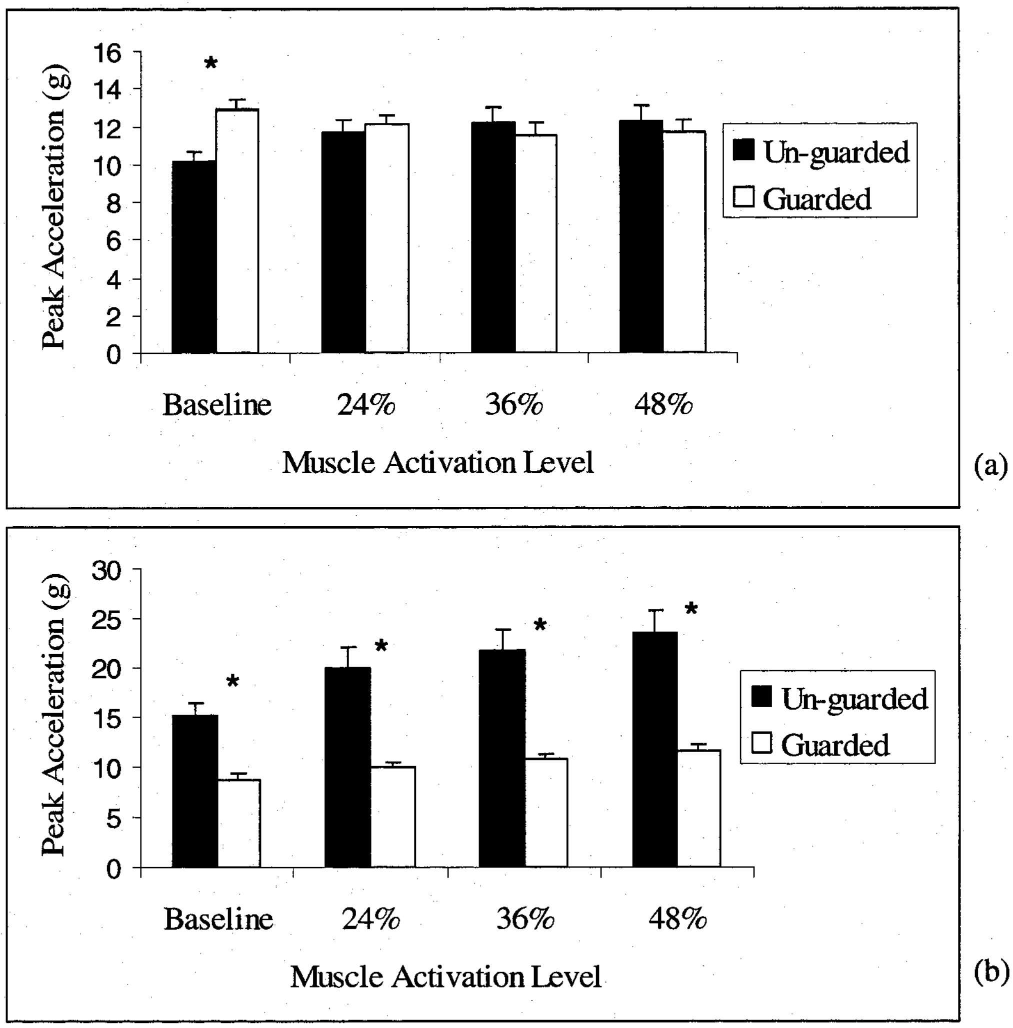


Figure 20. Wrist guard by muscle activation level interaction effect for peak axial (PA_{axial}) (a) and peak off-axis (PA_{off}) accelerations. (*$p<0.05$)

Acceleration Slope

AS_{axial} was significantly increased from 3193 g/s when un-guarded to 7715.7 g/s when guarded ($p<0.05$) (Figure 18b). Similar to the effect that wrist guard implementation had on the PA_{off}, AS_{off} was also significantly decreased by over half from 5642.4 g/s to 2606.8 g/s in the wrist guard condition ($p<0.05$). This main effect was also incorporated into a two-way wrist guard by muscle activation level interaction ($p<0.05$, $\omega^2 = 0.05$). AS_{off} in the un-guarded wrist condition increased from 3372 g/s at baseline to 7322 g/s at 48 % muscle activation. However, in the guarded wrist condition, AS_{off}

increased from only 1915 g/s to 3064 g/s at baseline and 48%, respectively. When guarded, the AS_{off} at the highest level of muscle activation was less than the AS_{off} at baseline in the un-guarded condition (Figure 21).

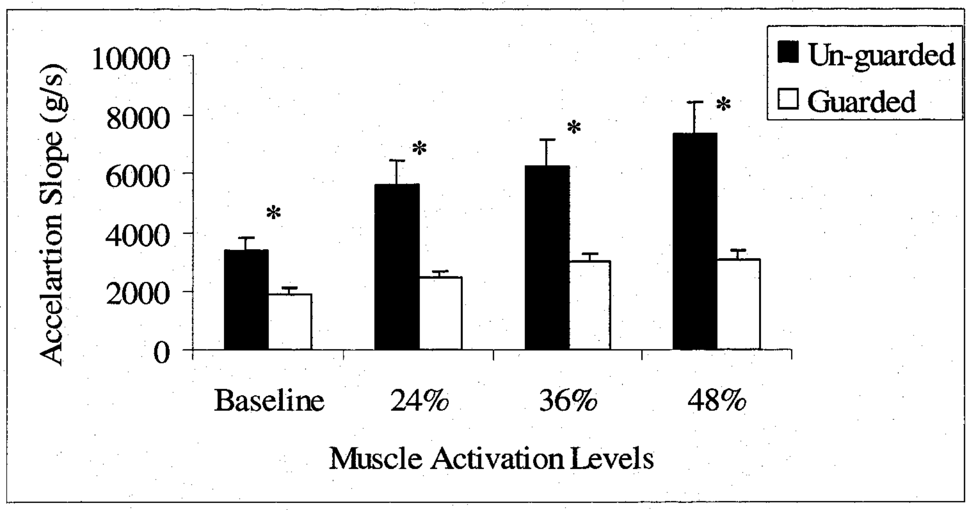


Figure 21. Wrist guard by activation interaction effects on off-axis acceleration slope (AS_{off}). (*$p<0.05$).

Time to Peak Acceleration

Wrist guards did not have a significant main effect on the mean TPA_{axial} (Figure 18c). However, a significant wrist guard main effect was observed for the TPA_{off} ($p<0.05$). When guarded, TPA_{off} was decreased on average by 15 % to 22 ms compared to 26 ms when the wrist was un-guarded. A significant two-way interaction was found between wrist guards and sex ($p<0.05$, $\omega^2 = 0.25$). TPA_{off} in males was relatively unaffected, decreasing by only 1 ms compared to females who experienced a 7 ms decrease in TPA_{off} when the wrist was guarded.

Elbow Angle

On average, subjects were able to maintain the static elbow angles at the moment of impact (Figure 22). Once impact occurred, approximately 12° of post impact flexion took place in both the straight arm and normal elbow conditions.

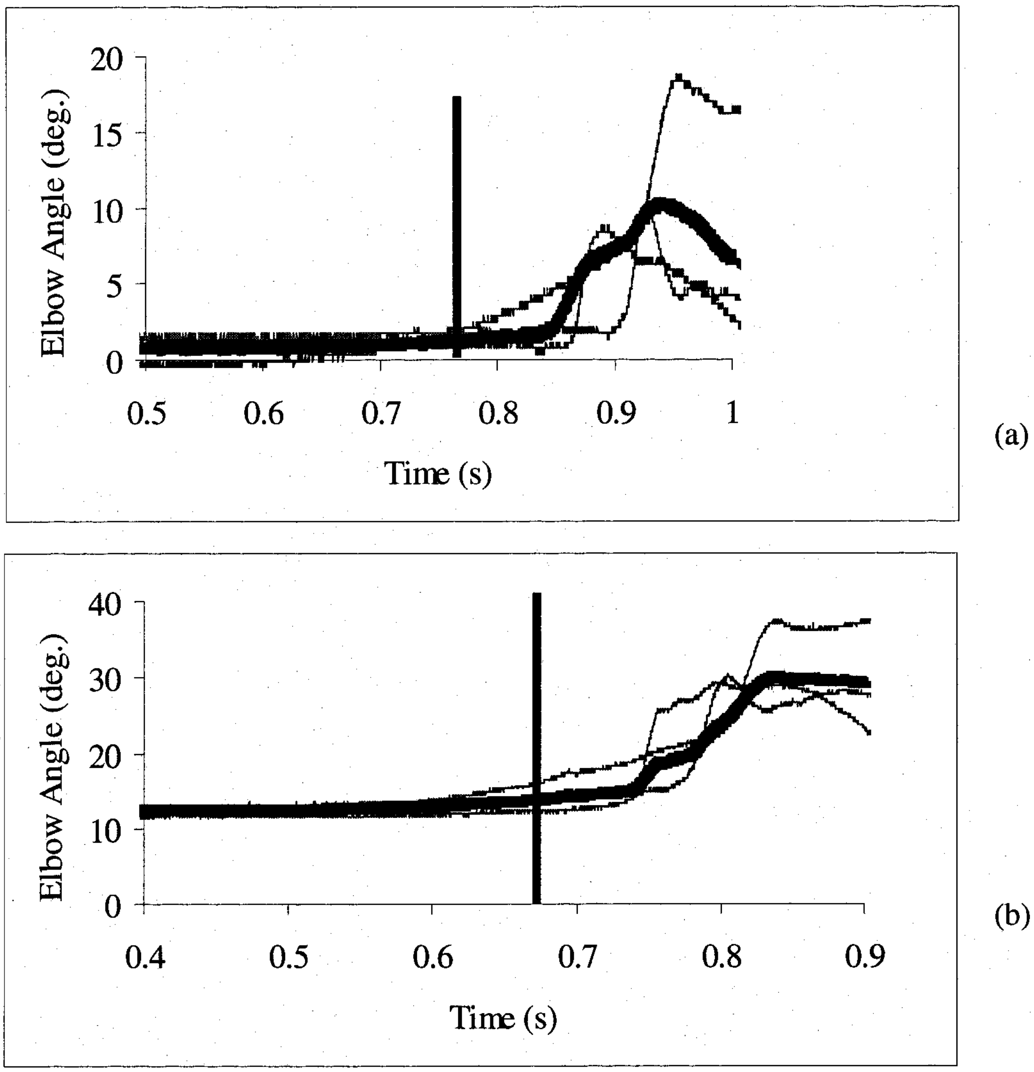


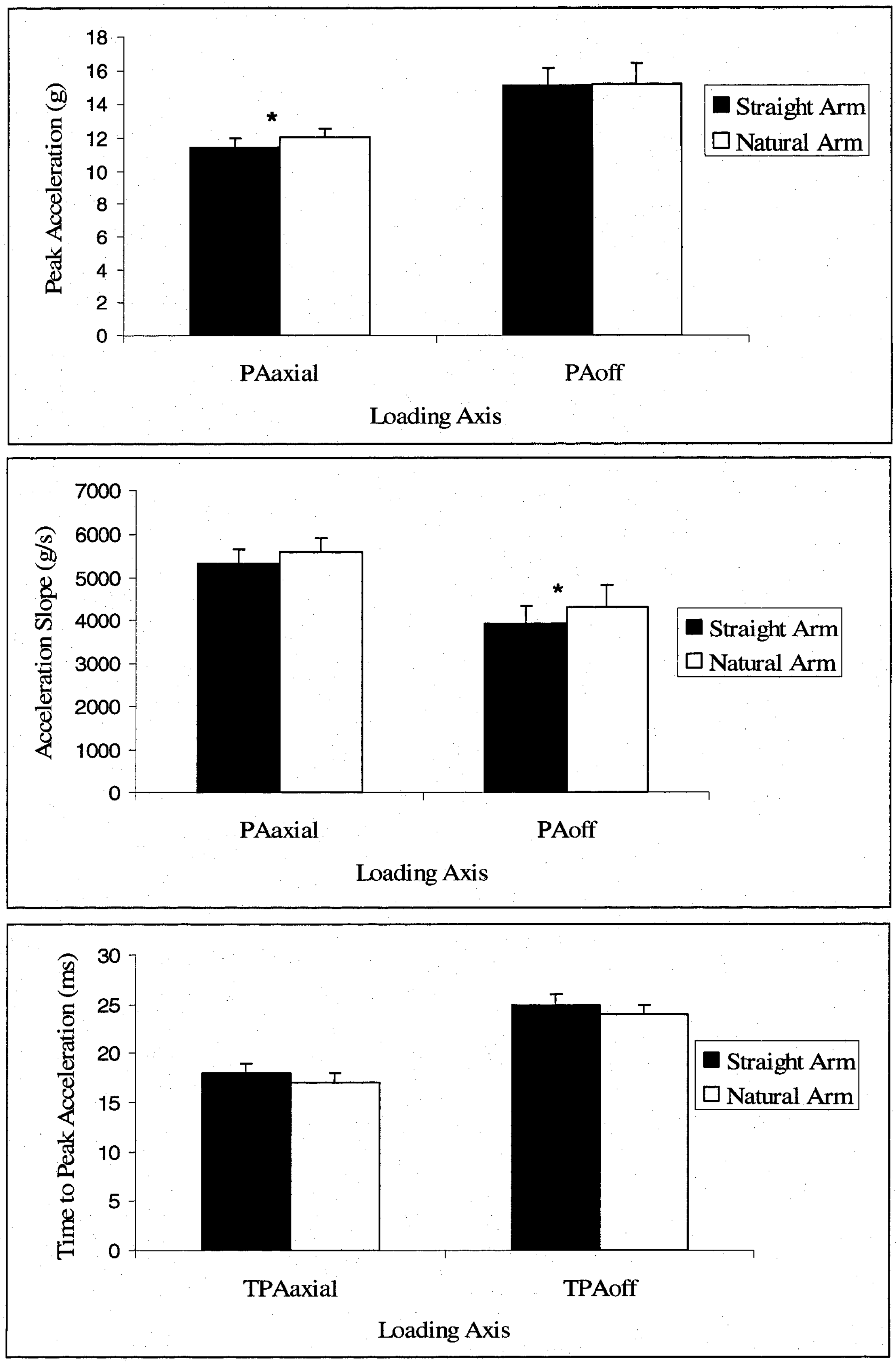
Figure 22. Straight arm elbow angles (a) and natural arm elbow angles (b) throughout impact. The moment of impact is represented by the vertical black line. Three different trials are represented by thin lines. Mean elbow angle is shown as the thick black line.

Peak Acceleration and Time to Peak Acceleration

Elbow angle significantly affected the PA_{axial} response of the wrist, with mean values of 11.5 g and 12.1 g in the straight and normal arm conditions, respectively ($p<0.05$) (Figure 23a). Elbow angle did not have a significant effect on PA_{off} or on TPA_{axial} and TPA_{off} (Figure 23c).

Acceleration Slope

AS_{axial} increased when the arm was impacted in a natural posture, but this change was not significant. However, AS_{off} increased significantly on average from 3926 g/s in the straight arm condition to 4323.3g/s when the subjects were impacted with natural arm postures ($p<0.05$) (Figure 23b).



(a)

(b)

Figure 23. Mean axial and off-axis peak accelerations (PA_{axial} & PA_{off}) (a),

(c)

acceleration slopes (AS_{axial} & AS_{off}) (b) and time to peak axial (TPA_{axial}) and off-axis (TPA_{off}) accelerations (c) for the straight and natural elbow angle conditions (*$p<0.05$).

Muscle Activation

Peak Acceleration

Similar to the effects of the wrist guard, the PA response at the wrist, as a function of muscle activation level, was highly dependent on the direction of loading. PA_{axial} was not significantly affected by changes in percentage of muscle activation. There was less than a 0.5 g difference between the mean baseline level of muscle activation (~12% MVE on average) and the mean at the 48 % level. However, PA_{off} was significantly affected by changes in muscle activation, resulting in a 6.0 g increase from baseline to 48%. PA_{off} was also significantly lower at baseline muscle activation than at the 24 % and 36 % levels ($p<0.05$) (Figure 24a).

Acceleration Slope

Significant differences in mean AS_{axial} values were found between the baseline and 48 % muscle activation levels ($p<0.05$). AS_{axial} increased linearly from 4745 g/s at baseline to 6829 g/s at 48 %, with 48 % muscle activation also significantly greater than both the 24 % and 36% muscle activation levels. Similarly, the mean AS_{off} value at 48 % (5193 g/s) was also significantly greater than at the baseline level (2643 g/s) ($p<0.05$) (Figure 24b).

Time to Peak Acceleration

TPA_{axial} and TPA_{off} both decreased significantly at the wrist. It only took about 36 % of the time to reach the peak acceleration at the 48% activation level than it did at baseline for both TPA_{axial} (16 ms from 25 ms) and TPA_{off} (21 ms from 33 ms) ($p<0.05$) (Figure 24c).

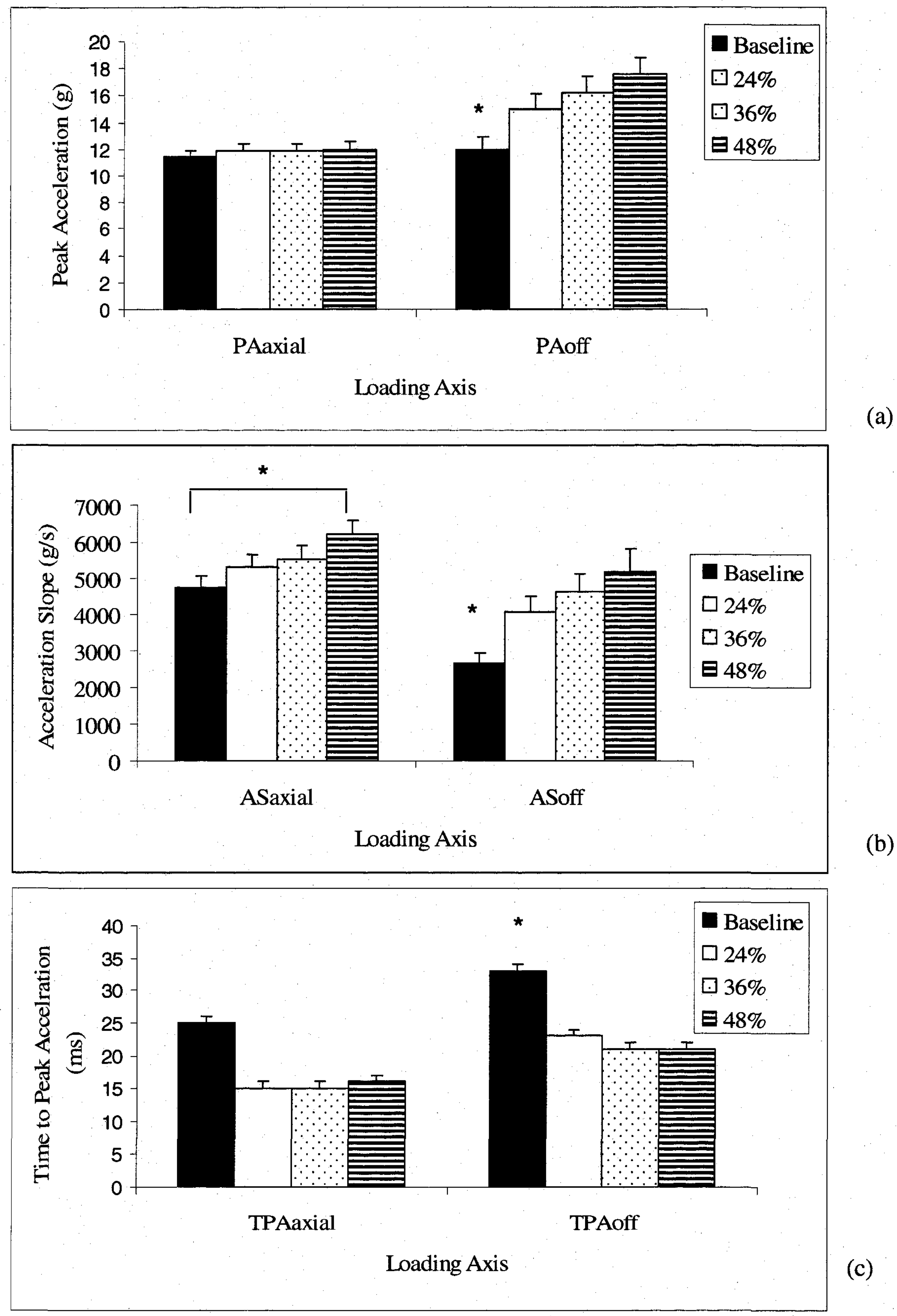
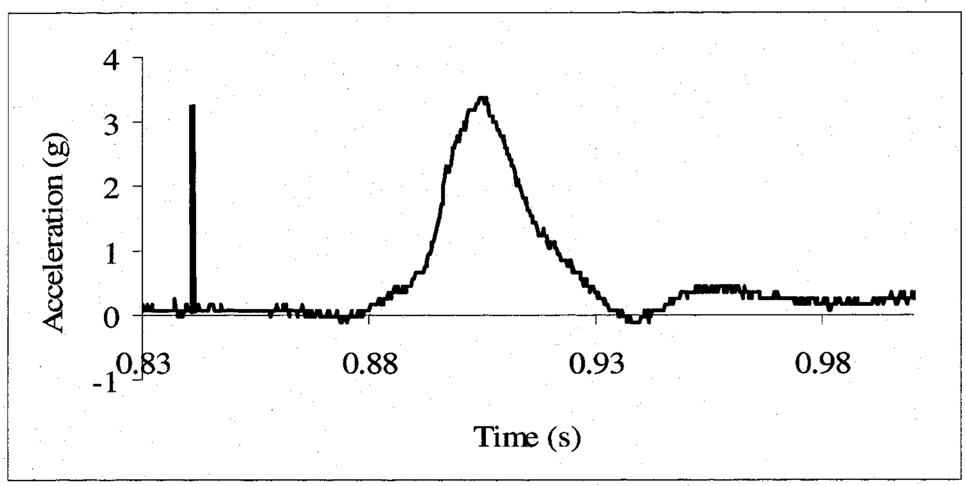


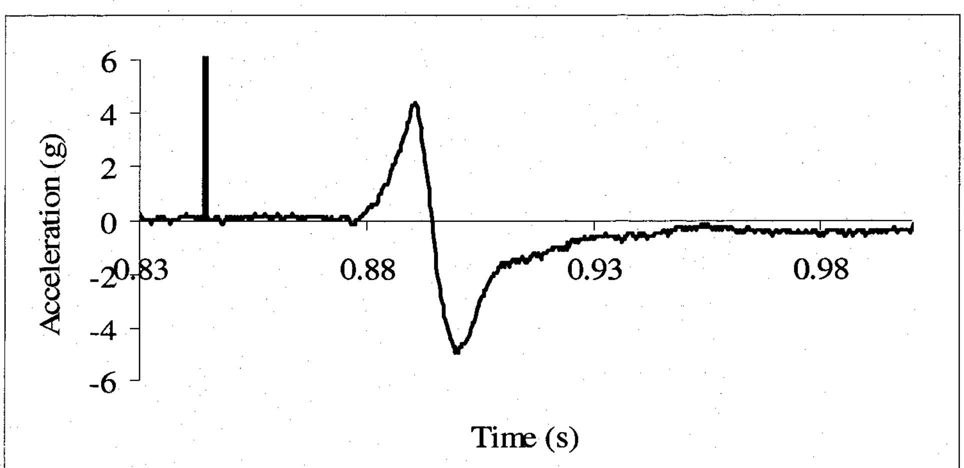
Figure 24. Mean peak axial (PA_{axial}) and off-axis (PA_{off}) accelerations (a), acceleration slopes (AS_{axial} & AS_{off}) (b), and time to peak acceleration for axial (TPA_{axial}) and off-axis (TPA_{off}) (c) at the wrist for all muscle activation levels (* $p<0.05$).

Elbow Response

The magnitude of the mean PA_{axial} at the elbow was approximately one third of that at the wrist (i.e. 4.78 g). The mean (SD) PA_{off} at the elbow was minimal, at only 1.0 g (0.6) (Table 4). The elbow was loaded at much lower rates than the wrist, with mean (SD) acceleration slopes of 612 g/s (72) and 331 g/s (160) in the axial and off-axis directions, respectively. The time to peak acceleration in the axial direction occurred approximately 22 ms (0.01) after impact, while TPA_{off} occurred earlier at 16 ms (0.01) on average post impact (Figure 25).



(a)



(b)

Figure 25. Actual axial (a) and off-axis (b) acceleration traces measured at the elbow. The solid vertical line represents hand/force plate impact.

Table 4. Mean (SD) elbow acceleration responses. (*$p<0.05$, [a] $p<0.05$ with all levels of muscle activation, [b]$p<0.05$ with baseline level of muscle activation).

		Sex		Guard		Elbow Angle		Muscle Activation			
Acceleration Response	Overall	Female	Male	Un-guarded	Guarded	Straight	Natural	Baseline	24%	36%	48%
PA_{axial} (g)	4.8	4.7	4.9	5.7*	3.8	4.9	4.7	4.5_a	4.8	4.9	5.0
	(0.3)	(0.4)	(0.4)	(0.4)	(0.5)	(0.3)	(0.3)	(0.3)	(0.3)	(0.3)	(0.3)
PA_{off} (g)	1.8	0.9	1.0	2.8*	-0.8	-1.7*	3.7	1.5	1.4	0.9	0.1[a]
	(0.6)	(0.9)	(0.9)	(0.9)	(0.6)	(0.8)	(0.7)	(0.6)	(0.7)	(0.8)	(0.7)
AS_{axial} (g/s)	612	648	577	881*	344	575	650	522[a]	625	630	673
	(72)	(104)	(97)	(113)	(36)	(68)	(79)	(61)	(80)	(80)	(76)
AS_{off} (g/s)	331	430	231	627*	35	-421*	1083	351	405	315	253
	(160)	(234)	(219)	(260)	(14)	(193)	(212)	(112)	(160)	(216)	(217)
TPA_{axial} (ms)	22	23	20	20*	24	23*	21	25	21[b]	20[b]	23
	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(2)	(1)	(1)
TPA_{off} (ms)	15	15	16	13*	18	18*	14	18	15[b]	15[b]	16
	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)	(1)

Wrist Guards

Peak Acceleration

Unlike the response at the wrist, wrist guards were found to have a significant main effect on all of the dependent variables at the elbow (Figure 26). PA_{axial} was significantly decreased from 5.7 g in the un-guarded condition to 3.8 g when the wrist was protected. The wrist guard main effect was incorporated into a higher order two-way wrist guard by sex interaction ($p<0.05$, $\omega^2 = 0.179$). Although males and females both experienced a decrease in PA_{axial}, the difference was greater for males than for females (3.6 g vs. 1.4 g, respectively). When un-guarded, PA_{axial} was significantly less in females than males, but when the wrist was guarded, PA_{axial} was significantly less in males (Figure 27).

Wrist guards also had a significant affect on PA_{off}, both in magnitude and direction. When un-guarded, the acceleration response occurred in the positive direction at 2.8 g. However, the wrist guard caused the peak of the impact to change direction and decrease in magnitude to 0.8 g. The PA_{off} wrist guard main effect was incorporated into a two-way interaction with elbow angle ($p<0.05$, $\omega^2 =0.127$). When the arm was straight, both the un-guarded and guarded conditions produced negatively directed peak off-axis accelerations. However, when the upper extremity was impacted in the bent arm position, the PA_{off} in the guarded wrist remained relatively low at just 0.7 g. The mean PA_{off} experienced in the un-guarded wrist condition increased to 6.7 g.

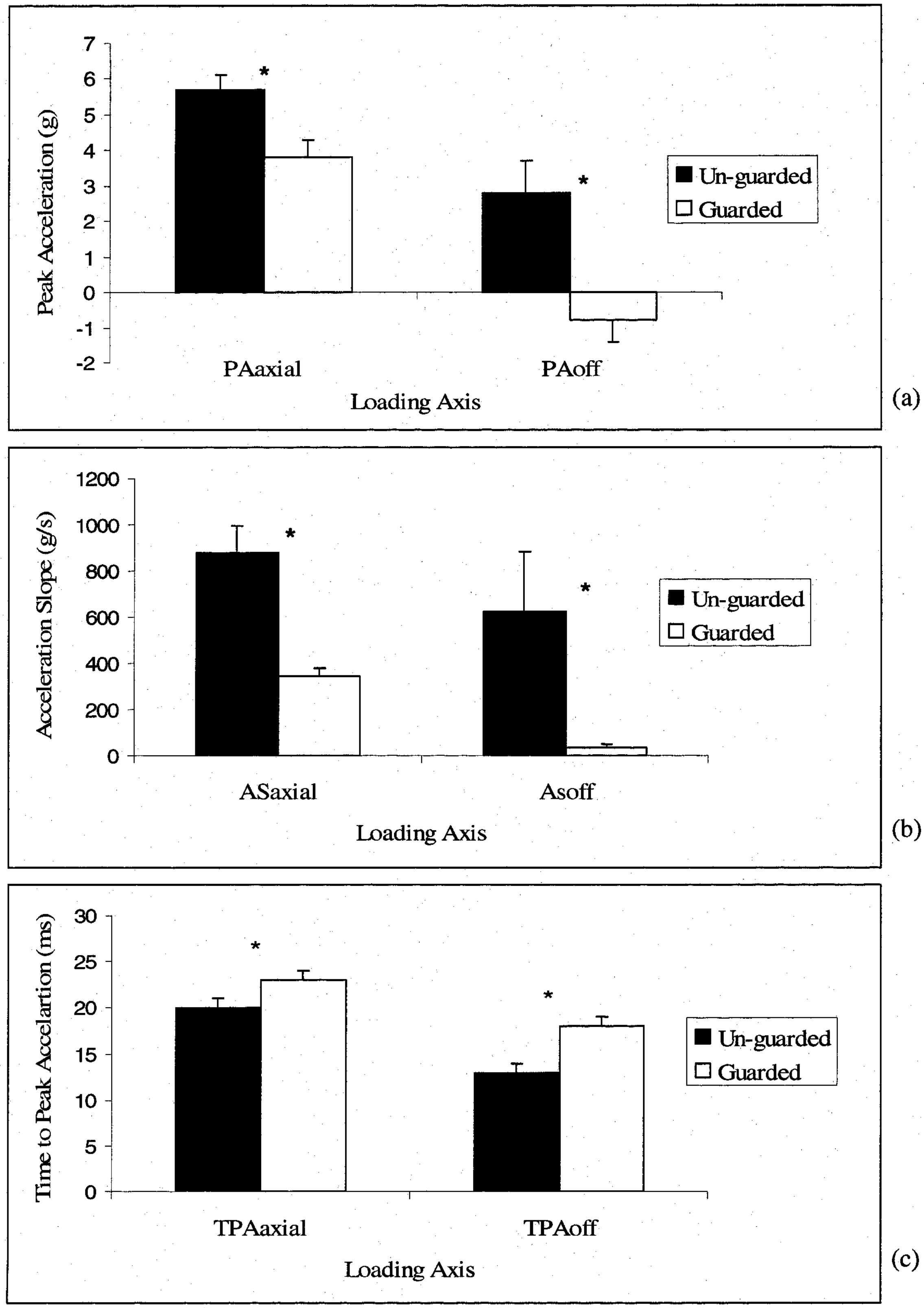


Figure 26. Mean axial and off-axis peak accelerations (PA_{axial} & PA_{off}) (a), axial and off axis acceleration slopes (AS_{axial} & AS_{off}) (b) and time to peak axial (TPA_{axial}) and off-axis (TPA_{off}) acceleration (c) at the elbow for the un-guarded and guarded conditions (*$p<0.05$).

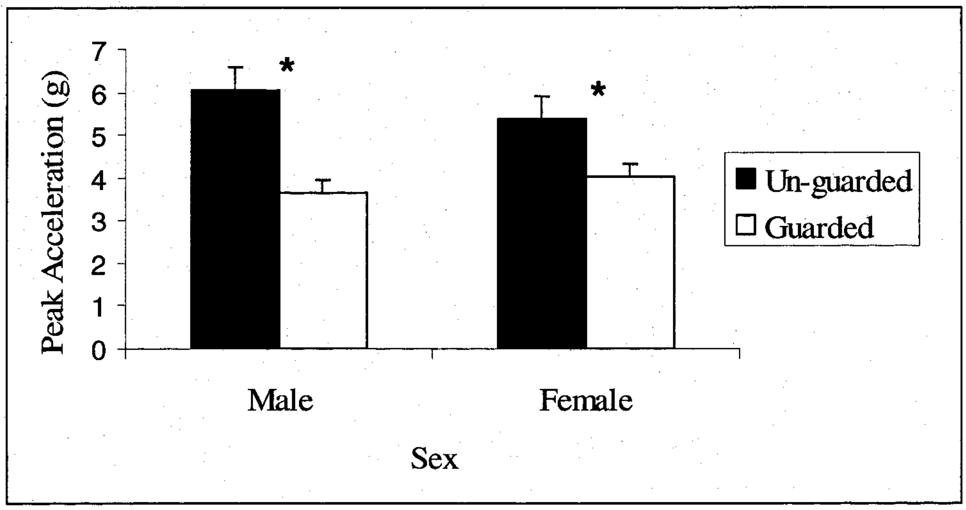


Figure 27. Wrist guard by sex interaction effect measured at the elbow (*p<0.05).

Acceleration Slope

A significant decrease in AS_{axial} was also noted between the guarded and un-guarded conditions (Figure 26b). When guarded, the elbow was loaded at than 40% of the rate than without the guard (881 g/s vs. 344 g/s, respectively). AS_{axial} was incorporated into a higher order two-way interaction with muscle activation ($p<0.05$, $\omega^2 = 0.06$). Un-guarded, AS_{axial} increased from 723 g/s at baseline levels to 976 g/s at 48 % muscle activation. However, AS_{axial} stayed relatively constant across all levels of muscle activation during the wrist guard conditions (Figure 28).

Similarly, the mean AS_{off} was decreased by a factor of twenty on average when the guards were used ($p<0.05$) (Figure 26 b). AS_{off} was also significantly affected by a two-way, wrist guard by elbow angle interaction ($p<0.05$, $\omega^2 = 0.36$). In both the un-guarded and guarded wrist, the change in elbow angle created a change in loading direction. However, when un-guarded AS_{off}, increased from – 600 g/s when the arm was straight to 1800 g/s during the natural arm impacts. AS_{off} increased to only 300 g/s when the arm was bent from -200 g/s in a straight arm impact when the wrist guard was used (Figure 29).

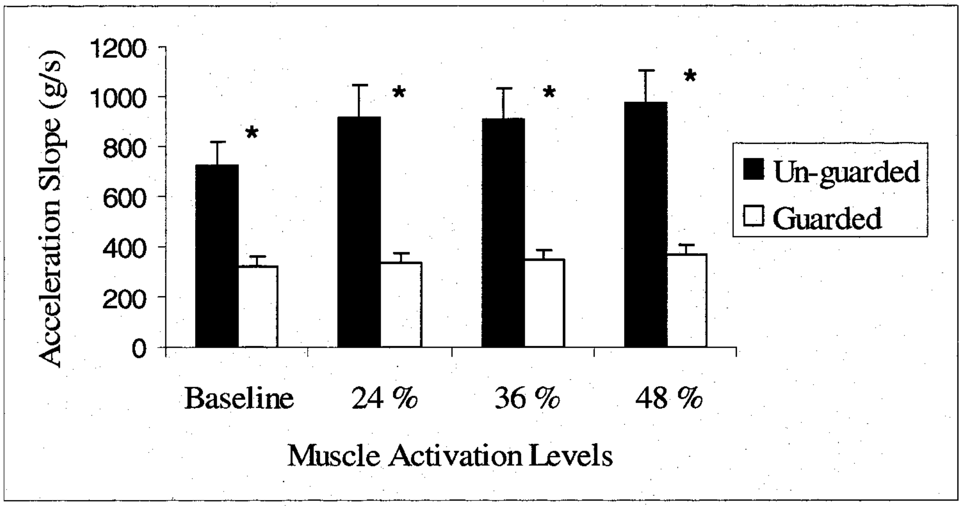


Figure 28. Wrist guard by muscle activation interaction effects on AS_{axial}, measured at the elbow (*p<0.05).

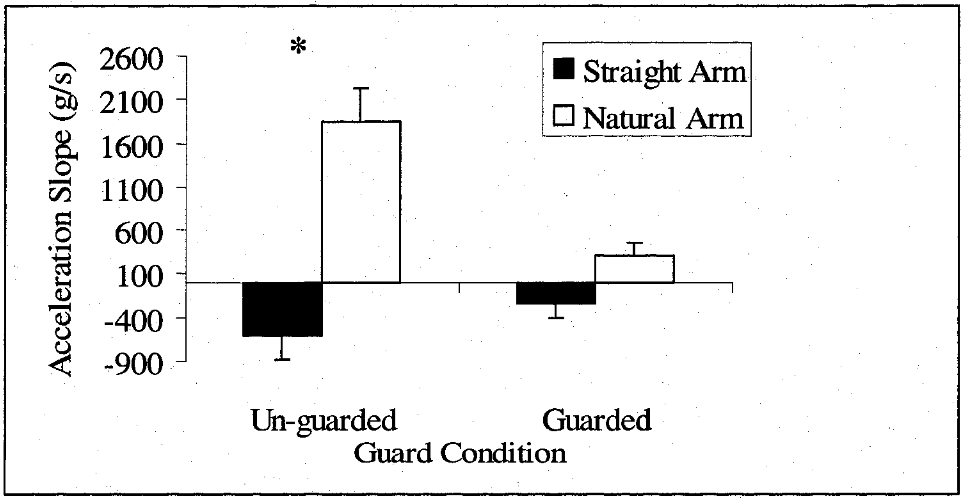


Figure 29. Elbow angle by wrist guard interaction effect on AS_{off} measured at the elbow (*p<0.05).

Time to Peak Acceleration

Mean TPA_{axial} values increased significantly to 24 ms when the wrist was guarded, from 20 ms in an un-guarded wrist ($p<0.05$). Significant increases of similar magnitude occurred in the TPA_{off}, changing from 13 ms to 18 ms in the un-guarded and guarded conditions, respectively (Figure 26c)

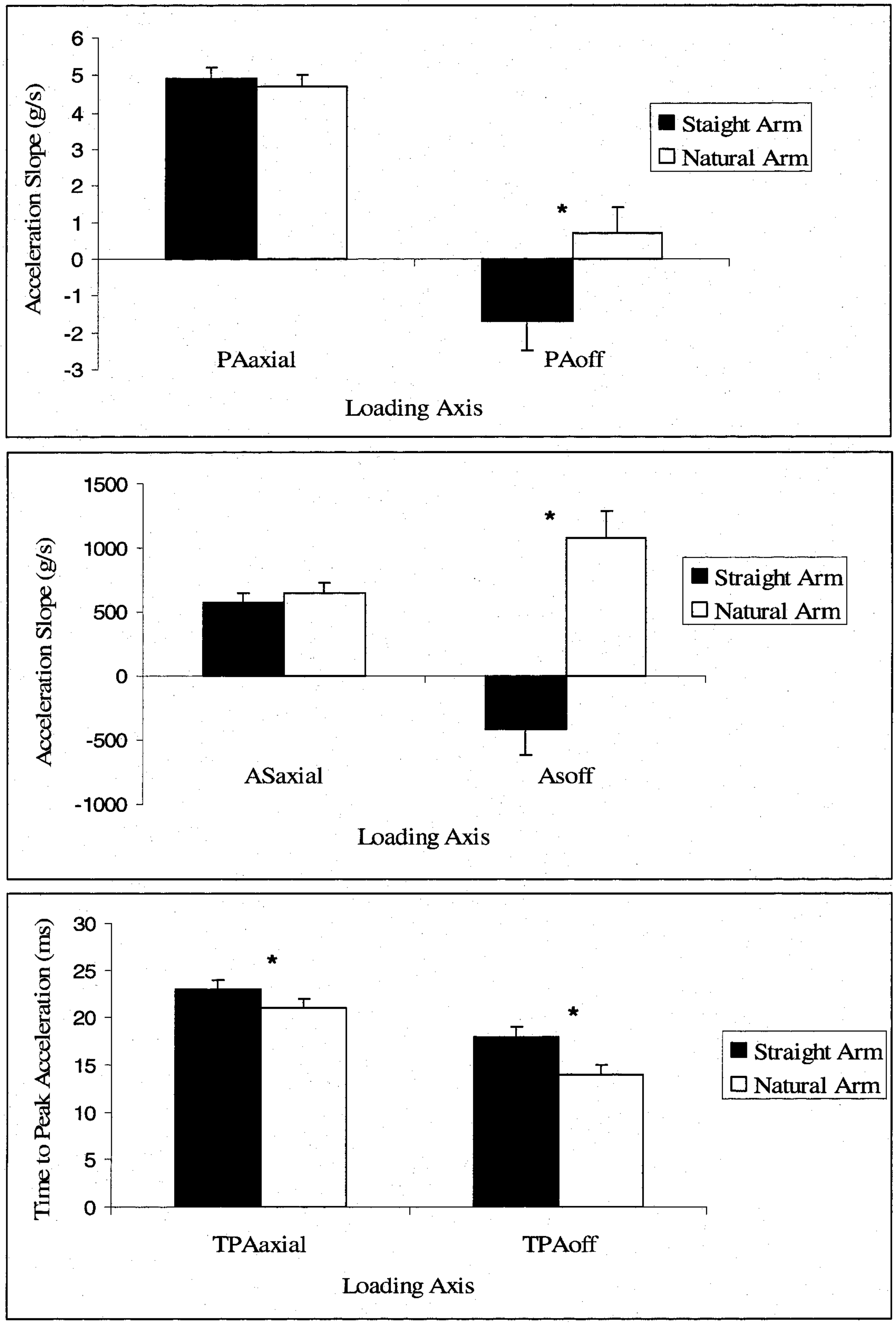
Elbow Angle

Peak Acceleration and Acceleration Slope

Changes in elbow angle from the straight arm impacts to the natural impacts, did not significantly affect either the PA_{axial} or AS_{axial}. However, as was suggested above, PA_{off} and AS_{off} were similarly affected in both magnitude and direction by the two different elbow angles ($p<0.05$) (Figure 30). Straight arm impacts resulted in a PA_{off} of -1.8 g, which is representative of a superiorly directed acceleration. However, natural arm impacts resulted in a significant increase in the magnitude of the mean PA_{off} to 3.7 g, which occurred in the opposite direction than in the natural arm condition.

Time to peak Acceleration

Changes in elbow angle from a straight arm impact to a bent arm impact led to significant decreases in both TPA_{axial} and TPA_{off}. TPA_{axial} decreased from 23 ms to 21 ms while TPA_{off} decreased from 18 ms to 14 ms ($p<0.05$).



(a)

(b)

(c)

Figure 30. Mean axial and off-axis peak accelerations (PA_{axial} & PA_{off}) (a), acceleration slopes (AS_{axial} & AS_{off}) (b) and time to peak axial (TPA_{axial}) and off-axis (TPA_{off}) acceleration (c) measured at the elbow for the straight arm and natural arm conditions (*$p<0.05$).

Muscle Activation

Peak Acceleration

Mean PA_{axial} was significantly increased ($p<0.05$) from 4.5 g to 5.0 g at the elbow when muscle activation was increased from the baseline level of approximately 12 % to the highest level of 48 % MVE. Differences were also significant between the baseline and the 24 %, and the 36 % activation levels. The elbow responded differently in the off-axis, with PA_{off} decreasing significantly between the baseline (1.5 g) and 48 % muscle activation (0.1 g) levels (Figure 31a).

Acceleration Slope

AS_{axial} significantly increased between the baseline and the three other levels of muscle activation ($p<0.05$). The largest difference occurred between the baseline level and 24 %, where it increased by 16 % from 522 g/s to 624 g/s, and eventually reached 673 g/s at 48 %. There were no significant differences for the AS_{off} (Figure 31b)

Time to Peak Acceleration

The mean TPA_{axial} values decreased significantly between baseline and 24 %, as well as between the baseline and 36 %. However, at 48 %, TPA_{axial} increased to 23 ms, a time similar to the baseline TPA, and differed significantly only with the TPA_{axial} at 36 % muscle activation. Similar to TPA_{axial}, TPA_{off} increased significantly to 15 ms at the 24 % and 3 6 % levels, and then decreased at 48 % to 16 ms, which was not significantly different than the values at any other activation level (Figure 31c).

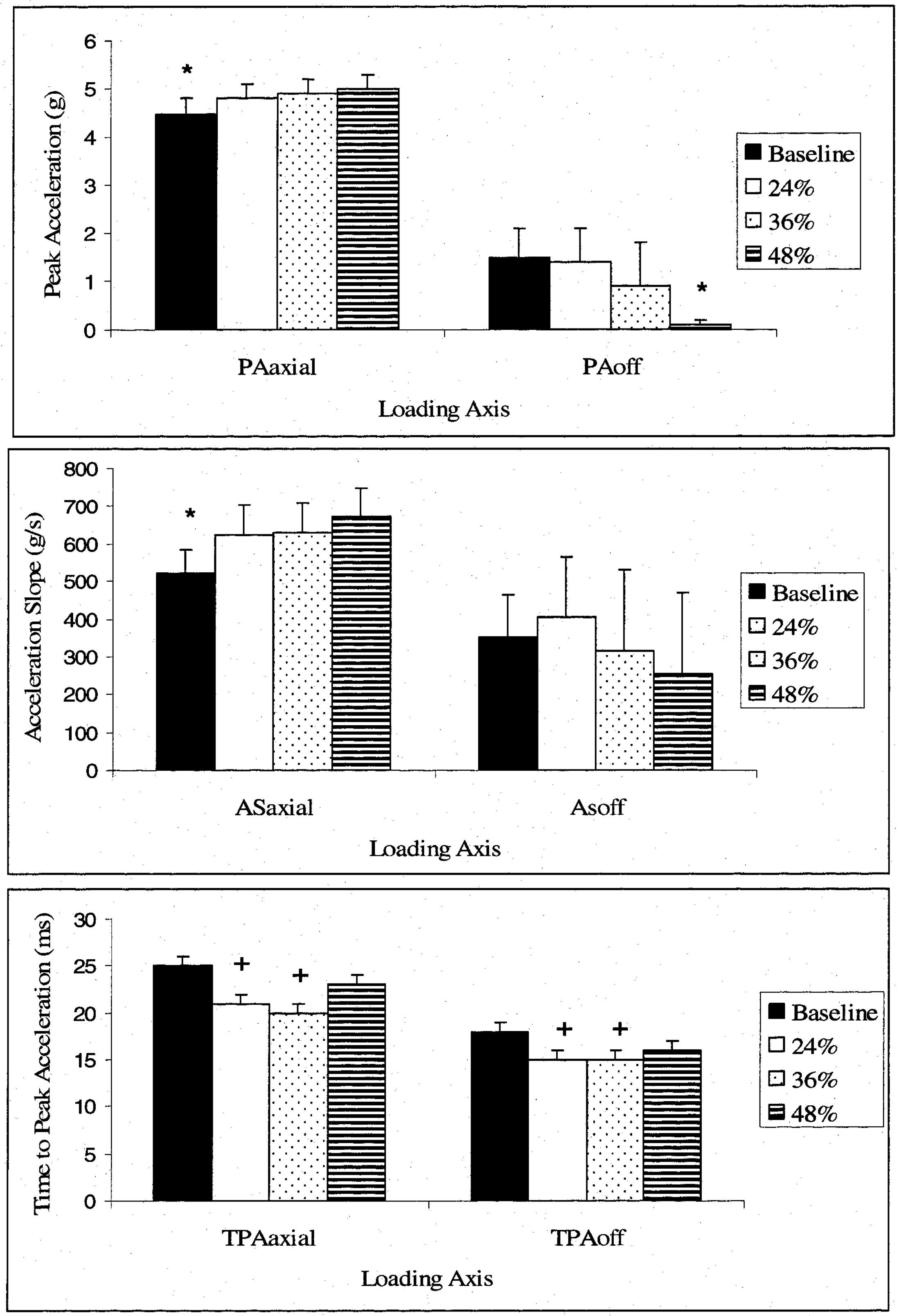


Figure 31. Mean axial and off-axis peak accelerations (PA_{axial} & PA_{off}) (a), acceleration slopes (AS_{axial} & AS_{off}) (b) and time to peak axial (TPA_{axial}) and off-axis (TPA_{off}) acceleration for the different levels of muscle activation (*p<0.05). + represents a significant difference with baseline muscle activation.

Chapter V
Discussion

Overview

To date, this is the first study to quantify the acceleration parameters in the upper extremity of living people in response to impact loading. Active and passive force attenuating mechanisms were tested under impact conditions that were designed to simulate the forces applied to the upper extremities following a forward fall. The attenuation response of the upper extremity to the tested mechanisms was found to be highly dependent on the joint location (wrist vs. elbow) as well as loading axis (axial vs. off-axis; see section 3.3.5).

Between the wrist and the elbow there was an overall decrease in the PA_{axial} response of 59 %. Wrist guards did not significantly decrease the PA_{axial} at the wrist. However, PA_{off} and AS_{off} at the wrist, as well as all acceleration response parameters at the elbow, were significantly decreased when the wrist was guarded.

Elbow angle significantly increased the PA_{axial} at the wrist, despite the decrease in the IRF measured at hand/force platform interface. At the elbow, the off-axis acceleration responses were shown to significantly increase during the natural arm impacts.

The increased stiffness associated with increasing muscle activation levels did not affect the axial response at the wrist, while PA_{off}, AS_{off} and TPA_{off} were all significantly increased from baseline to 48 % muscle activation. The elbow response clearly illustrates the disabling of the force attenuating characteristics of the soft tissue; as muscle activation levels increase from baseline to 48 %, all acceleration response parameters were increased, while the time to peak acceleration was decreased along both axes.

Kinematic impact data, including impact reaction force and velocity were compared to previous data (Chiu & Robinovitch, 1998; DeGoede & Ashton-Miller, 2002), to ensure that subjects were impacting at physiologically safe levels, as well as to validate the relatively novel impact method employed here. The results for this study display the same trends reported elsewhere with regards to elbow angle and wrist guard effects (DeGoede & Ashton-Miller, 2002, Hwang et al. 2006), and muscle activation effects (Pain & Challis, 2002) on IRF.

The reliability of the methods used here was controlled for by ensuring that all data collection equipment was calibrated and only one researcher was responsible for instrumenting all of the subjects. Also, subjects were impacted three times in each condition and the average of the three impacts was used for further statistical analysis. The IRF data were analyzed after each impact to ensure that the forces did not exceed the targeted parameters, and velocity was maintained at 1.0 m/s for all impacts.

Impact Force, Velocity and Kinetic Energy

The two handed impact approach that was used in the current study produced impact forces that were half the magnitude as those reported by Chiu & Robinovitch (1998) at similar impact velocities (0.2-0.8 m/s). Therefore, impact velocity was increased to 1 m/s so that the IRFs applied to the right hand more accurately matched the forces that occur during falls from relatively low heights, as reported previously (Chiu & Robinovitch, 1998).

Peak impact reaction forces (IRF) measured at the hand/force plate interface varied by approximately 10% (134 N) between the left and right hands, and on average, the right hand impacted 0.4 ms before the left. Although the majority of the subjects tested (n=22) reported being right hand dominant, there was no significant difference between the left and right hand dominant subjects for the hand that produced the greatest force. The results also indicated that handedness did not determine which hand impacted first. Subjects were instructed to impact the force plates with the left and right hand simultaneously and were visually monitored to ensure that there was no obvious or intentional bias for one hand over the other. However, subjects may have been more aware of the right hand given that only the right upper extremity was instrumented, and may have subconsciously led with the right hand more often.

Wrist guard use, elbow angle and muscle activation level had an effect on IRFs. Decreases in impact force between the guarded and unguarded conditions, as well as between flexed and straight arms, have been extensively reported throughout the literature (Hsiao & Robinovitch, 1998; DeGoede & Ashton-Miller, 2002; DeGoede & Ashton-Miller, 2003), and the results of this study support these findings in general. The dorsal strap of the wrist guard provided resistance as the subjects attempted to increase their ECU muscle activation levels, and therefore required less activation of the intrinsic musculature of the hand. Therefore, the upper extremity segment may have been less stiff when the wrist guard was implemented. However, to date, only one study (Pain & Challis, 2002) has reported on the effects of forearm muscle activation on peak impact forces. Pain & Challis (2002) reported that arbitrary increases in forearm tension increased the peak reaction force by approximately 30 %, similar to the 32 % increase in IRF reported in the current study, from 227.5 N at the baseline condition to 334.1 N when muscle activation was increased to 48 %. Furthermore, Nigg & Liu (1999) reported that simulated increases in segment stiffness of the lower extremity were associated with an increase in the ground reaction force following lower extremity impacts.

At impact, 41.7 J of energy was transferred to the upper extremities of subjects on average. Kinetic impact energy was unaffected by wrist guards or elbow angle, however, there was a 1.0 J difference in kinetic energy between the baseline and 48% muscle activation levels. Males produced an extra 6.0 J of energy compared to females. Since velocity was kept constant between trials and sexes, the higher impact energies for males resulted from them being 15 kg heavier than their female counterparts on average.

Electromyography

Subjects were able to accurately reach the desired levels of muscle activation, across all levels, with small differences between the target muscle activation level and the mean recorded level (maximum of 3 % difference). Also, the absence of changes in muscle activation during the wrist guard and elbow angle condition, suggests that the subjects were capable of maintaining the desired level of muscle activation as they approached impact.

There was little co-contraction of the FCU muscle group when the wrist guard was implemented or when impacts occur to a flexed arm. The increased FCU activation in response to increased ECU activity, may contribute to stiffening of the forearm structure. However, Potvin et al. (2001) reported a significant relationship between FCU and ECU activity during flexion tasks, and attributed it to a wrist stabilizing role. Given the relatively high levels of ECU muscle activation levels required of subjects in this study, it is possible that the FCUs were contributing to a stabilizing effect at the wrist.

Wrist Acceleration Response

Wrist Guard Effects

The protective force attenuating capacity of the wrist guard, at the wrist, was found to be dependent on the loading axis (axial vs. off-axis). PA_{axial} was generally unaffected by the use of a wrist guard. Kim et al. (2006) found similar results in the axial direction when they compared a bare hand to several different guarded conditions, including guards made from traditional plastic, sorbothane, air cells and air bladders. They reported that on average, 2.5 % more force was transmitted when one of the protective materials was being used compared to the bare hand.

However, even as muscle activation continued to increase, there was little change in the PA_{axial} and relatively little difference between unguarded and guarded conditions. This result suggests that although the segment is becoming increasing stiff, the wrist guard is absorbing, at minimum, a small proportion of the impact force. However, in contrast was the 50 % decrease in PA_{off} after the wrist guard was implemented in the current study. Also, AS_{axial} increased by 59 % (7700 g/s) when the wrist guard was in use, while the off-axis effects were reduced by almost 50 %.

These off-axis effects suggest that the volar splint was successful in absorbing the impact force and changing the loading axis. Instead of being absorbed through the hand, the impact force was being diverted down into the volar splint. However, the splint, being parallel with the long axis of the radius, simply transmitted the impact force from the distal to proximal forearm area (i.e. axially through the wrist) (Kim et al., 2006). The wrist guard by elbow angle and wrist guard by muscle activation interactions provide further evidence that the wrist guard was effective in absorbing, but not diverting the impact force axially. When un-guarded, the PA_{axial} increased from the straight arm to the natural arm conditions. However, when guarded, there was no difference between the two wrist guard conditions. A similar effect was noted in the muscle activation interaction. At the baseline levels, the PA_{axial} was greater when the wrist was guarded. However, as muscle activation was increased, PA_{axial} between the un-guarded and guarded condition became relatively similar, and although insignificant, the PA_{axial} at 48% was lower in the guarded condition.

An extended wrist at impact has been identified as a risk factor for wrist injuries (Hwang et al., 2006; Troy & Grabiner, 2006). However, when the hand is constrained within the wrist guard at impact, the wrist is unable to extend to the same degree as it would when it is unguarded. Recently, Troy & Grabiner (2007) suggested that the loading axis has a significant influence on the risk of a radial fracture. They reported that it requires a smaller off-axis impact force to initiate an injury than a force that occurs along the axis of the segment. When loaded axially, the fracture force was increased to 2830 N from 1448 N when the impact force was applied to the off-axis (Troy & Grabiner, 2007). Not only were the off-axis accelerations in this study shown to be greater than the axial accelerations, but it was also shown that wrist guards may be a suitable mechanism for protection against relatively harmful off-axis forces.

The wrist guard was also shown to protect the off-axis responses (PA_{off}, AS_{off}, and TPA_{off}) against the effects of increasing muscle activation. With the wrist constrained within the guard, thereby providing external resistance of the wrist guard strapping across the dorsum of the hand, subjects required much less effort to reach the higher levels of muscle activation. Therefore, muscle activation levels could be achieved with minimal activation of the palmar musculature.

Elbow Angle Effects

Although IRFs were significantly decreased by flexing the forearm about the elbow, PA_{axial} was significantly increased. AS_{axial} increased by approximately 200 g/s when the arm was flexed, however, this increase was not significant. In contrast, AS_{off} was found to increase significantly by approximately 20%.

The "natural angle" of 12° that was used in this study has been shown to occur naturally when people impact the ground and try to protect themselves with outstretched arms (Hsiao & Robinovitch, 1998; DeGoede & Ashton-Miller, 2003; Lo, et al., 2003). Changing the elbow angle is considered to be an active force attenuation mechanism because IRFs have been shown to decrease as a result (DeGoede et al., 2002 & Lo et al., 2003). However, the response of the wrist in this study contradicts previously reported results. The natural elbow angle of 12° may not have been a large enough change from the straight arm condition to be effective in reducing the impact effects at the wrist. Also, in an attempt to control elbow angle across all subjects, impacts occurred to a statically flexed arm. This negates much of the active force absorption effects that are experienced when the elbow is allowed to flex freely at impact. However, the elbows continued to flex by approximately 12-15° post impact, and after the PA_{axial} at the elbow was achieved. Furthermore, keeping the wrist angle constant between the straight arm and natural arm conditions required the hand to impact lower on the force plate during natural arm impacts. Subsequently, shoulder flexion was decreased by approximately 5-10° to achieve the desired wrist and elbow postures.

Two theories may help to explain the increase in acceleration response at the elbow (both PA and AS) in the bent arm condition. The first draws on a motor control theory discussed by Medendorp et al. (2000). They suggest that, given the coupling of the shoulder and elbow joints, there are several strategies that can be drawn upon when positioning the arm, each producing a different forearm tilt in oblique arm configurations. It is possible in the current study, that the elbow-shoulder relationship created a forearm posture more susceptible to transmitting the impact force. Secondly, in lower extremity impacts, the hip plays a major role in the attenuation of impact force, mainly by hip flexion (Zhang, Bates & Dufek, 2000). The shoulder may be responsible for similar force attenuation when the upper extremity is impacted. During the straight arm impacts, the elbow has approximately 90° of range of motion available to assist in active force attenuation. However, only 80° of the range of motion is available when the arms are bent and the shoulders lowered, decreasing the ability of the shoulder to actively react to, and absorb the IRF.

The mean PA_{axial} for males did not change between the two elbow angles, whereas females, on average, experienced a small but significant 1.0 g increase in PA_{axial} when they impacted with flexed arms. In general, males tend to have increased upper extremity muscular strength (Miller, MacDonald, Tarnopolsky, & Sale, 1993). The increased strength allows males to maintain static elbow angles throughout impact compared to females who are more likely to bend the arms once impact occurs. DeGoede & Ashton-Miller (2002) also attribute sex differences to the differences in strength between males and females.

Muscle Activation Level Effects

Comparing the responses in the two loading directions provides insight into the role of increasing muscle activation levels (MALs) on the local wrist response. PA_{axial} was unaffected by changes in MALs, increasing by only 0.4 g from baseline to 48%. However, PA_{off} significantly increased by approximately 5.0 g from the lowest level of muscle activation to the highest. Interestingly, both AS_{axial} and AS_{off} were also found to increase significantly. The fact that AS_{axial} and not PA_{axial} were found to differ significantly can be described by the significant decrease in the TPA_{axial}. It appears that the increased segment stiffness associated with increased muscle activation (as suggested by Nigg & Liu, 1999 for the lower extremities for example) had little affect on PA_{axial} at the wrist. The bulk of the forearm muscle mass that would be responsible for increases in segment stiffness is located proximal to the wrist, approximately 2/3 of the way up the forearm. This mass is located too far away from the wrist to exert any type of passive axial force attenuating effect at the wrist. However, there are two possible explanations for both the significant increase in PA_{off} as well as AS_{axial} and AS_{off} as activation level increased. Although not directly measured, the intrinsic musculature of the hand (i.e lumbricals & palmar interossei) was also altered as subjects attempted to reach the higher levels of muscle activation by abducting and extending their fingers. The axial components of the acceleration response were minimally affected given the location of the extensor tendons and palmar soft tissue, which stiffen the pathway through the hand, as opposed to down the segment.

Elbow Acceleration Response

In general, there was a 59% decrease in PAaxial, and 88 % decrease in ASaxial and TPAaxial increased by 18 % from the wrist to the elbow. The decrease in PAaxial is similar to the percent of force attenuation that was measured between the ankle and the knee of the lower extremity (Chu & Yazdan-Arkani, 1986).

Wrist Guard Effects

The wrist guard effects were found to be significant for all the dependent variables measured at the elbow. Axially, PA and AS were decreased by 33% and 61%, respectively, in the guarded condition. Peak off-axis accelerations not only decreased in amplitude, but also experienced a change in direction, on average. There was also a 94% decrease in the mean AS_{off} from 627 g/s to almost zero at 34 g/s. Although there was a significant difference found for the PA_{off}, it is suggested that the change in direction has little functional significance given the relatively small magnitude of the PA_{axial} in the guarded wrist. This suggests that the wrist guard was effective in attenuating the impact force for both loading axes.

The wrist guard by elbow angle interactions further highlights the force attenuation characteristics of the wrist guard. When the wrist is guarded, the relatively large increases in PA_{off} and PA_{axial} that were experienced in the un-guarded wrist become substantially negated.

Males and females differed significantly in their response to the wrist guard. The PA_{axial} response was relatively similar in males and females. Males experienced a 50 % decrease in PA_{axial} when their wrists were guarded, while the female response was reduced by only 25 % on average. Two different sizes of wrist guards (large and small) were provided to ensure that each subject was fitted properly and comfortably. However, the surface area of the volar splint was 26 cm^2 larger in the large wrist guard than the small wrist guard. It was noted that males were typically fitted with the large wrist guards, while the majority of the females required the smaller guard. The force absorbing potential of any sports equipment is dependent on the surface area of the attenuating surface, and therefore may have played a differential role here.

Elbow Angle Effects

Similar to the axial response at the wrist, the flexed elbow posture did not decrease either the PA_{axial} or AS_{axial}. PA_{axial} remained virtually unchanged between the two elbow posture conditions, while the AS_{axial} increased by approximately 100 g/s when the elbow was flexed. It is suspected that the lack of force attenuation via elbow flexion can be explained by the same mechanisms as at the wrist. Although the global response is a decrease in the impact reaction at the force plate, the possible adoption of specific coordination patterns (Medendorp et al., 2000), combined with changes in shoulder flexion, likely caused the local impact responses at the wrist and elbow.

The off-axis response differed considerably from the axial response in that there was both a change in magnitude and direction for PA_{off} and AS_{off}. Negative accelerations were recorded during a straight arm impact for both PA_{off} and AS_{off}. On the other hand, when subjects were impacted with a bent arm, the acceleration response occurred in the positive direction and increased significantly from the straight arm condition, up to 3.71 g and 1083 g/s for PA_{off} and AS_{off}, respectively. The change in loading direction may be a result of the localized impact of the distal humerus within the trochlear notch. When a straight arm impact occurs, the distal humerus is inline with the long axis of the radius and ulna. Following the initial impact, and the positive PA, the negative axial acceleration may have been a result of the humerus impacting the ulna along the long axis. However, when the arm was flexed, the trochlea of the humerus would have rotated in the sagittal plane within the trochlear notch, and would have become more aligned with the off-axis of the elbow.

The off-axis (positive) loading of the elbow in the bent arm condition would occur as the trochlea of the humerus is impacted within the trochlear notch. This would result when the hand and forearm come to rest while the remainder of the body's mass is arrested following impact. The TPA_{off} occurred in just 15 ms, given the short distance between the two anatomical structures. This response helps to explain the many different types of elbow fractures reported in the literature (Ring et al., 1997; Houshian et al., 2001; Wake et al., 2004; Doornberg, & Ring, 2006). Wake et al. (2004) reported different fracture sites in the elbow based on elbow angles under static compressive loading (Figure 32). The elbow angle results reported here for dynamic loading, support the findings of Wake et al. (2004). As elbow angle changed from 0° to 12°, different anatomical structures are potentially impacted.

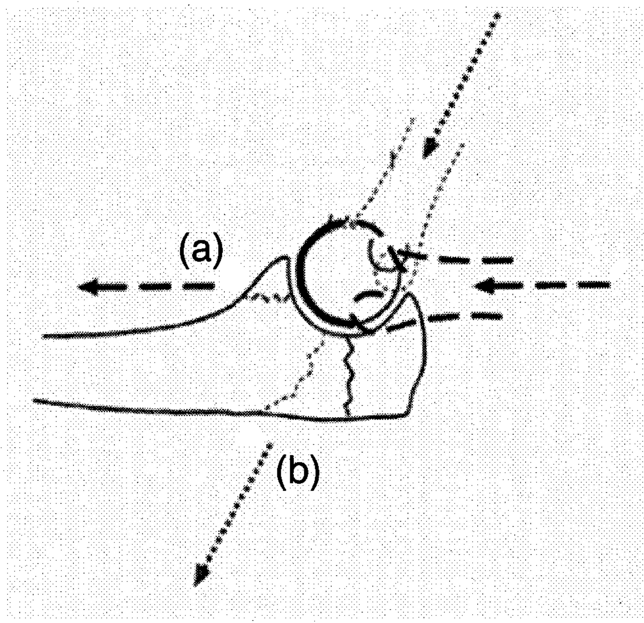


Figure 32. Schematic representation of the effects of different elbow angles on the loading axis at the elbow. Fracture of coronoid process results from a straight arm impact (a), and fractures to the olecranon occur from bent arm impacts (Wake et al., 2004).

Muscle Activation Effects

As the muscle activation levels increased from baseline to 48 %, the axial acceleration responses were also significantly increased. Baseline muscle activation levels were approximately 15 % MVE for all subjects, an activation level that represented the effort needed to sustain relaxed wrist extension throughout the impact. The 48 % muscle activation level required subjects to isometrically contract the forearm extensors, and in many cases, represented the maximum un-resisted effort of the subjects. According to Derrick, Hamill, & Caldwell (1998), Liu & Nigg (1999), and Wakeling, Liphardt & Nigg (2003), as the muscular activation in the leg was increased, so to was the stiffness. This change in stiffness subsequently changed the force attenuating capabilities of the segment. Applying this to the upper extremity, as the arm was impacted at baseline levels, there was minimal stiffness in the segment and the soft tissue absorbed and attenuated the impact force to a greater degree. When the segment became maximally stiff (i.e. at 48% activation in this study), impact forces were transmitted to a larger extent by the soft tissues. Further highlighting the active shock transmission capabilities of the forearm muscular was the increase in AS_{axial} from baseline to 48 %. The forearm was axially loaded by an additional 151 g/s from baseline to 48 % muscle activation, agreeing with lower extremity acceleration slope data presented by Holmes & Andrews (2006) and Flynn et al. (2004).

The elbow PA_{off} also responded significantly to changes in muscle activation. However, these results contrasted those in the axial direction. PA_{off} and AS_{off} decreased by 91 % and 28 %, respectively. The large decrease in PA_{off} is evidence that the active soft tissues of the forearm segment are providing more of a benefit than just force attenuation in the off-axis. The extensor muscles of the forearm originate to some extent, on the humerus, and cross over the elbow joint before inserting onto the metacarpals and phalanges of the hand. When these muscles are contracted, they act to increase the stabilization of the elbow joint by compressing the irregularly shaped joint surfaces against one another (Dunning, Zarour, Pastterson, Johnson & King 2001; Safran & Bailargeon, 2005). With the joint in a more stabilized position, prior to and throughout the impact, the head of the humerus is less likely to displace and impact the ulna. This finding suggests that there is an optimal level of muscle activation that translates into minimal force transmission and maximal joint stability, a combination that minimizes the risk of elbow injuries.

Limitations

The impact apparatus that was used in this study required subjects to remain in a seated position while they travelled towards the impact surface, similar to DeGoede et al. (2002). Although this orientation is not entirely representative of the positioning which occurs naturally during forward falls, the subjects safety was a main concern, and the seated human pendulum was designed to meet the basic postural demands of a falling human, in terms of the trunk, shoulder and elbow angles at impact (Hsiao & Robinovitch 1998; Lo et al., 2003). Furthermore, the impact parameters of force and velocity remained within physiological limits of the subjects.

The subjects in this study were all healthy Human Kinetics students. The BMI data suggest however, that this sample (mean BMI = 25 kg/m^2) is comparable to 58.5 % of the Canadian population with a BMI in the range of 20 kg/m^2 – 27 kg/m^2 (Statistics Canada, 2001). Furthermore, given the novel approach of this research, only healthy, non-pathological individuals were studied. Future research should be conducted to determine the effect of soft tissue and bone pathologies on force attenuation.

The palmar soft tissue, similar to the calcaneal fat pad, may also be capable of active and passive impact force attenuation. However, the current study did not take this into account. There is no method currently available that allows researchers to accurately measure the mass of palmar soft tissue in-vivo, and collecting muscle activation data on the intrinsic musculature of the hand would prove difficult in an impacting scenario.

In an attempt to accommodate all subjects two sizes of wrist guards were made available (small ad large). Although every subject fit into one of these wrist guards, some individuals required more tension (i.e. to keep the wrist guard in place for smaller subjects) or less tension (i.e. larger subjects) in the strapping, therefore the level of tension applied to the back of the hand and distal forearm was not constant across subjects.

One of the most noticeable limitations of this study was the placement and type of accelerometers used to measure acceleration responses. Kim et al. (1993) & Lafortune et al. (1995) have shown that transient force waves are more accurately measured directly at the bone via bone mounted accelerometers. However, given the invasive nature of this technique, skin mounted accelerometers were used in the collection of acceleration responses at the wrist and elbow. Although skin mounted accelerometers are more prone to picking up movement of the underlying soft tissue, precautions were taken to ensure minimal interruption. The accelerometers were adequately pre-loaded onto the underlying bony structures; a fact that was indicated by the visible superficial tissue markings after the accelerometers were removed, and the slight discomfort that subjects felt during the testing procedures as the accelerometers were forced into the underlying tissue.

Chapter VI
Conclusions

The current study simulated a forward fall impact to the upper extremities to asses the force attenuating characteristics of wrist guards, changing elbow angles, and increasing muscle activation levels. The following conclusions can be drawn:

Wrist Guards

At the wrist, the H_0 for PA_{axial}, and TPA_{axial} could not be rejected. The H_0 for PA_{off}, AS_{axial}, AS_{off}, and TPA_{off} was rejected. At the elbow, the H_0 could be rejected for all acceleration responses. Therefore, it can be concluded that wrist guards are effective in reducing acceleration responses at the elbow, and off-axis acceleration effects at the wrist. However, there is little advantage to using wrist guards to protect the wrist from axially applied accelerations.

Elbow Angle

At the wrist, the H_0 for PA_{off}, AS_{axial}, TPA_{axial}, and TPA_{off} could not be rejected. The H_0 for PA_{axial} and AS_{off} could be rejected. At the elbow, the H_0 for PA_{axial} and AS_{axial} could not be rejected. However, the H_0 for the PA_{off}, AS_{off}, TPA_{axial} and TPA_{off} could be rejected. Therefore, it can be concluded that, although impacting with a natural arm angle decreased the IRF, a flexed elbow angle does not help reduce the acceleration response at the wrist.

Muscle Activation level

At the wrist, the H_0 for the PA_{axial} could not be rejected. However, the H_0 for the PA_{off}, AS_{axial}, AS_{off}, TPA_{axial}, and TPA_{off} could be rejected. At the elbow, the H_0 for the AS_{off} could not be rejected, but the H_0 for the PA_{off}, PA_{axial}, AS_{axial}, TPA_{axial} and TPA_{off} could be rejected. Therefore, it can be concluded that increasing muscle activation from baseline to 48 % increased the off-axis response at the wrist and the axial elbow response, and decreased the off-axis elbow response.

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