# On the Load-Displacement Behavior of the Tensed Adult Upper Extremity under Impulsive End-Loads: Theoretical and Experimental Studies of Age and Gender Effects

by

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**To my whole family**

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## Preface

The chapters have been written as separate manuscripts for submission, and there may be some reputation between chapters, specifically in the materials and methods sections.  Chapter 2 and 5 have been published in *Annals of Biomedical Engineering*.

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# ABSTRACT

The upper extremities are often used to protect the head and thorax by bracing for impact, particularly in falls to the ground. The impulsive loads they impose on the hand and wrist can be substantial, exceeding one body-weight. If the upper extremity then “gives way” or flexion buckles at the elbow then a head injury is likely, particularly in the elderly; but if the elbows are fully extended to prevent buckling, then the risk for wrist fracture increases. A current knowledge gap includes the biomechanical factors that determine the threshold load required to flexion-buckle the elbow of an end-loaded and pretensed human upper extremity.

In this thesis we use computer simulations and *in vivo* experiments to explore how age, gender, initial elbow angle, arm muscle strength and pre-contraction level and lumped contractile properties about a joint affect upper extremity deflection under impulsive end-loading. The experimental results show that gender and age affect the rotational stiffness and damping coefficients of muscles acting about the elbow and shoulder when estimated by dynamic optimization. The pre-contraction levels of arm and shoulder muscles significantly affected these coefficients. Computer simulations predict that advancing age, female gender and insufficient arm and shoulder muscle pre-contraction level adversely affect upper extremity buckling loads. Kinetic, kinematic and myoelectric studies suggest the speed of propagation of the impulsive load along the upper extremity is such that arm and shoulder muscles must be pretensed prior to impact:

no neuromuscular reflex is rapid enough to increase arm muscle tensile stiffness to prevent flexion buckling. Pre-contraction level and gender significantly affected the rate of propagation of an impulse along the upper extremity.

The findings provide a framework for better understanding how biomechanical factors determine whether or not an arm will buckle when end-loaded during a fall arrest. We conclude that in order to help safely arrest falls older women and men need to avoid using hyperextended arms when possible, use an adequate pre-contraction level in the arm muscles to prevent buckling, and maintain as much arm protraction strength as possible, perhaps most conveniently by regular push-up exercises.

# CHAPTER 1

# INTRODUCTION

## 1.1 Epidemiology of Falls and Fall-Related Injuries

### 1.1.1 Falls and Fall-related Injuries as an Important Socioeconomic Problem

Falls are a leading cause of unintentional injury in all ages (for example, CDC 2009-2018 and WISQARS$^{TM}$ (Web-based Injury Statistics Query and Reporting System)). Falls can be particularly devastating for older adults (CDC 2009; Davis et al., 2010; Ambrose et al., 2013). For example, more elderly die from falls than die from motor vehicle accidents in the United States (Binder, 2002) and falls are the leading cause of injury-related deaths among adults aged over 65 years (WISQARS$^{TM}$). Furthermore this is a growing problem because the rate of fall-related deaths among older adults in the United States has risen significantly over the past decade (WISQARS$^{TM}$).

In terms of the scale of the problem, each year in the United States, nearly one-third of older adults experience a fall (Hornbrook et al., 1994; Hausdorff et al., 2001; Stevens et al., 2006) posing a major threat to health and independence. For example, about one out of ten falls among older adults result in a serious injury, such as a hip fracture or head injury, that requires hospitalization. In addition to the physical and emotional pain, many people need to spend at least a year recovering in a long-term care facility and it often has a substantial effect on a person's self-confidence, mobility, independence, and quality of life (Tinetti et al., 1988; Evans et al. 2001; Oliver et al. 2004; Terroso et al., 2013).

The socioeconomic costs of falls are considerable. By 2020, the annual direct and indirect costs of fall-related injuries including cost associated with lost work due to injury or death are expected to reach $54.9 billion, in 2007 dollars (Englander et al., 1996). The total costs of fall-related injuries are highest between the age of 25 and 44 years because of the morbidity costs ($8.3 billion) associated with restricted activity level and productivity lost (Rice and MacKenzie, 1989). Among children aged less than 15 years, falls are the leading cause of an emergency department visits, accounting for an estimated 2.2 million visits in 2006 (WISQARS™). Infants and children who fall from low heights are at substantial risk for head injuries, and those falling from heights of over 10 feet can also sustain multiple serious injuries. The total direct cost of fall injuries among adults aged over the age of 65 years in 2000 was $19 billion (Stevens et al., 2006). So interventions that reduce fall-related injuries can potentially have a marked socioeconomic benefit at any age.

### 1.1.2 Age and Gender Differences in Falls

The number of falls increases progressively with age in both genders and all racial and ethnic groups (Fuller, 2000; Ambrose et al., 2013). Women are 67% more likely than men to have a nonfatal fall injury, while men are more likely to die from a fall than women. Rates of fall-related fractures among older adults are more than twice as high for women as for men (Stevens, 2005). Schultz et al. (1997) reviewed the literature on findings from large and carefully-executed studies of rates of fall and fall-related injuries among older adults. The studies of rate of falls, that involved over 4,500 adults 65 or more years old, found those rates to range from 137 to 690 falls per 1,000 persons per year, with older females falling from 1.3 to 2.2 times more often than older males. The studies of rates of fall injuries requiring medical attention, conducted among over 38,000 adults, found, for example, that fall injuries leading to hospital admission or death occurred in males and females at rates per 1,000 persons per year of 1.88 and 0.83 for those of ages 20-29 years, and increased steadily with age to 6.97 and 15.58 for those of ages 70-79 years. Correspondingly, female/male ratios of these serious fall injury rates increased with age from 0.44 to 2.24.

The gender difference in the rate of fall and fall-related injuries is present regardless of country. For example, among persons aged 65 years and older in Australia, the number of fall-related injury admissions to care hospitals (per 100,000 population) was 6,383 for men and 15,306 for women in 2008/2009 (Watson and Mitchell, 2011). Age-adjusted hip fracture rates per 10,000 population in United Sates were 291.6 for men and 510.9 for women aged 65 years and older in 2006 (Stevens and Rudd, 2010). Why older females are twice as susceptible to fall-related injuries as older males has partly been explained by higher prevalence of osteoporosis which is known to lower the fracture toughness of bone (Seeman, 2001). However, it has been suggested that the biomechanics of the fall can be more important determinants of fracture than bone density (Stevens, 2010), so this partially motivates many of the studies in this dissertation.

### 1.1.3 Fall-Related Injuries

Major injuries, including head trauma, soft tissue injuries, fractures and dislocations, occur in 5-15 percent of falls in any given year (Fuller, 2000; Terroso et al., 2013). Fractures account for 75 percent of serious injuries. Hip and wrist fractures are costly, particularly in older women (Cummings et al., 1985; Terroso et al., 2013; Ambrose 2013). Most hip fractures occur in older adults and over 90 percent of hip fractures in older adult results from a fall (Grisso et al., 1991; Goldacre et al., 2002; Ambrose et al., 2013; Terroso et al., 2013). The sequelae from hip fractures can be devastating. Approximately 20 percent of women who fracture their hip do not survive the first year after fracture, while another 20 percent do not regain the ability to walk unassisted (Schneider & Guralnik 1990). Over 85 percent of wrist fractures involve falls. The incidence of wrist fracture rises from age 50 to 65 years and then reaches a plateau after age 65 years (Melton III et al., 1988; Kristinsdottir et al., 2001), but why this occurs is not yet understood.

### 1.1.4 Fall Direction and Impact Site

Fall direction and impact site are important factors affecting injury risk and type. O'Neill et al., (1994) found 60% of falls in older adults are forward, 20% were to the side, and 20% were backward; similar numbers have been found by others (e.g., Vellas et al., 1998). Men are more likely to fall to the side, and women are more likely to fall forward or trip (O'Neill et al., 1994; Berg et al., 1997). These data underlie the rationale for comparing fall arrest behaviors of men and women in the forward direction. Gait speed and the type of disturbance are known to affect fall direction with faster gait speeds associated with forward falls and slower gait speed associated with lateral falls (Smeesters et al., 2001). Lateral falls, slips and impact at the hip are associated with hip injury; forward falls, trips and impact with the hands are associated with upper-extremity injury (Hayes et al., 1993; Nevitt and Cummings 1993; Cumming and Klineberg 1994; Nyberg et al., 1996; Palvanen et al., 2000).

Individuals of any age often use the upper extremities to help arrest a fall, presumably to reduce risk of injury to the head or torso. But falls directly onto the hands with fully extended upper extremities cause a high rate of wrist fractures (Nevitt and Cummings 1993; Palvanen et al., 2000). So it would be logical to avoid the use of fully extended arms in this way. But the behavior of slightly flexed upper extremities subjected to this type of impulsive end loading is not well understood and is a focus of the present dissertation. Because falls are a leading cause of TBI (traumatic brain injuries), particularly over the age of 70 years (Ingebrigtsen et al., 1998; Kannus et al., 2001), the protection afforded by upper extremities in these cases was clearly ineffective. So even though protective use of the hands is associated with a potential risk for wrist injury, the risk-benefit ratio seems reasonable given the potential severity of head or hip injury in the absence of such a strategy.

Vellas et al. (1998) demonstrated that the part of the body receiving the main impact in a fall was, in order of frequency: the hand (50% males, 33% females) and buttock (18% males, 24% females), followed by the head, knee, and arm. In older women, the most common fall-related fracture sites are the upper extremity, the hip, and the trunk or neck in that order (Sattin et al., 1990). For older men a similar pattern is observed although the fracture rates are halved. It is noteworthy that the first impact is with a hand when falling to the ground (O'Neill et al., 1994). Furthermore, although women seem to use their hands less often for arresting falls than men, the rate of upper extremity fracture is much higher in women than men (Sattin et al., 1990). It could be that the older women fail to use their arms as effectively as older men to arrest falls, either by choice or otherwise. Indeed, it has been suggested that reduced upper extremity strength may render the arms ineffective in arresting falls to the ground (Nevitt and Cummings, 1993; Palvanen et al., 2000; DeGoede and Ashton-Miller, 2003). Clearly, use of the upper extremities is important for arresting many falls, but a better understanding is needed of the biomechanical factors that affect the efficacy of upper extremity use in such falls.

## 1.2 Biomechanical Factors of Falls and Fall-Related Injuries

Biomechanical capacities such as strength, power, range of motion and coordination deteriorate with advancing aging and with disease. Changes occur in myoelectric latencies (Cuccurullo, 2004; Verdú et al, 2000; Norris et al., 1953), reaction time (Pijnappels et al., 2010), the afferent system (Jimenez-Andrade et al., 2012), proprioception (Wingert et al., 2013), muscular strengths (Daly et al., 2013) and the rate of development of those strengths (Thelen et al., 1996). These biomechanical factors are important for falls because they affect response time, reaction time, activation time of muscle, muscle and joint properties (strength, stiffness, damping) of lower extremities, upper extremities, hip and trunk, use of arm to arrest a fall, use of body joints to alter the impact severity, and use of multiple impact sites to arrest a fall (for example, Lo and Ashton-Miller, 2008).

Epidemiological studies have shown the arms are used to arrest falls with considerable morbidity and 96% of distal forearm fractures are caused by falls, likely due to the attempt to arrest the fall with the upper extremities (Keegan et al., 2004), presumably to reduce risk of injury to the head or torso noted above. DeGoede et al. (2003) showed that the biomechanics of how a fall is arrested are important in determining the loads applied to the upper extremity musculoskeletal system.

### 1.2.1 Impulsive Forces and Moments on the Arm during a Fall

Fall arrests can be thought of as having two phases after fall initiation: a pre impact and an impact phase. The interval from loss of balance to impact constitutes the pre-impact or descent phase which can last up to seven tenths of second from a loss of balance (Hsiao and Robinovitch, 1998). During the actual impact, the ground reaction force peaks within a few hundredths of a second after the hand contacts a surface or object (Dietz et al., 1981; Chiu and Robinovitch, 1998; DeGoede and Ashton-Miller, 2002; Lo et al., 2003). So, the impact phase duration lasts up to a few hundredths of second (Figure 1.1) or an order of magnitude shorter than the descent phase.

The impact phase duration is therefore so short that few, if any, volitional changes in neuromuscular state can be made during that phase. Hence the forces and moments produced by the impact are largely predetermined by the pre-impact neuromuscular state. That is, they are largely determined by the initial conditions under which the fall begins and the biomechanics of any neuromuscular response during the descent.

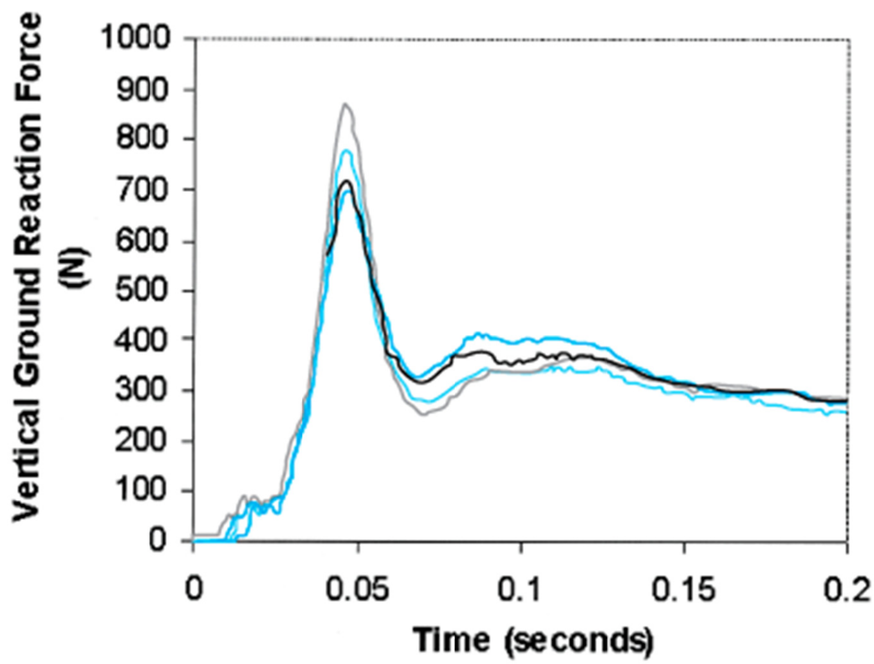


**Figure 1.1** Examples of the measured time history of the impact force on one wrist in four consecutive, sagittally symmetric, forward falls onto both hands performed by a healthy young male subject weighing 620 N. The subject was instructed to land naturally (DeGoede and Ashton-Miller, 2002).

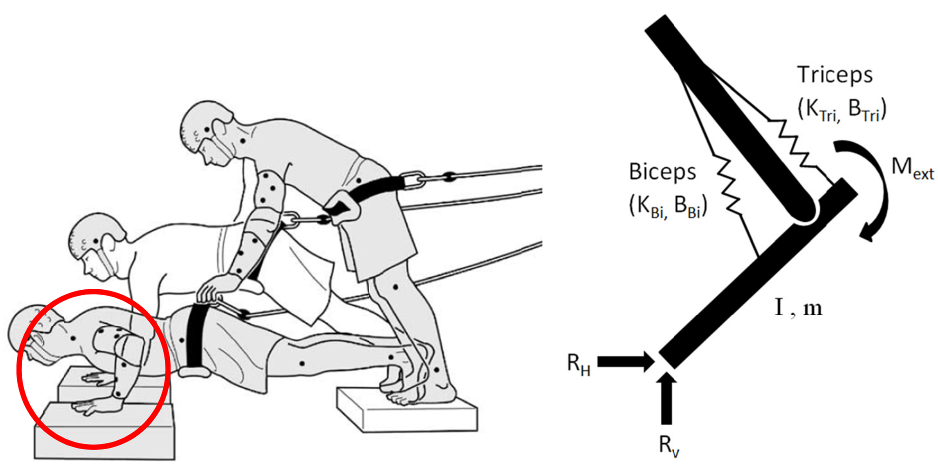


**Figure 1.2** Laboratory setup (left) of falls onto the arms (DeGoede and Ashton-Miller, 2002). The circle is the part of a forearm with the elbow joint and the wrist joint at impact. Schematic at right shows a free-body-diagram of a forearm having mass m and inertia I with rotational springs (K) and dampers (B) representing the elbow muscle actions in the sagittal plane, under external forces $R_H$ and $R_V$.

A free body diagram describes the forces acting about the elbow at impact (Figure 1.2). $M_{ext}$ is the external flexion moment of the elbow joint. $R_H$ and $R_V$ are ground reaction forces at the wrist joint in the horizontal and vertical directions. The segment mass, m, is located at the center of mass. I is the moment of inertia of the segment. The combined vector of $R_H$ and $R_V$ represents the ground reaction force, $R_F$ on the wrist. When $M_{ext}$ is applied at the elbow joint the joint dynamics can be modeled as

$$M_{ext} = I\ \ddot{\theta} + (B_{Tri} + B_{Bi})\dot{\theta} + (K_{Tri} + K_{Bi})\theta$$

Over the range of common comfortable walking speeds, approximately 0.5 to 1.5 m/s, walking speed has little effect on the kinetic energy that must be arrested (DeGoede, 2000). In terms of energy, however, the total energy of a fall while walking is larger than when falling and not walking because if one walks one adds the kinetic energy from walking to the potential energy. So one needs to dissipate more energy from a fall during gait than from the standing posture. This larger kinetic energy leads to a larger initial velocity on the upper body after a trip, so the impulsive moments acting at the wrist, the elbow and shoulder are increased in the case of a fall while walking or running. In other words, a fall from the standing posture such as when fainting may induce lower external moments on the upper body joints than when walking (Lo et al., 2003). Therefore a fall from walking or running imposes higher impulsive moments at the upper extremity joints.

At least two sets of factors can modulate the magnitude of the highest peak impact force (Figure 1.1): the mechanical properties of the impact surface and impacted soft tissues, and the kinematics of the body at and during impact. The latter factor has already been mentioned above in relation to walking speeds and joint moments related to impact (DeGoede, 2000).

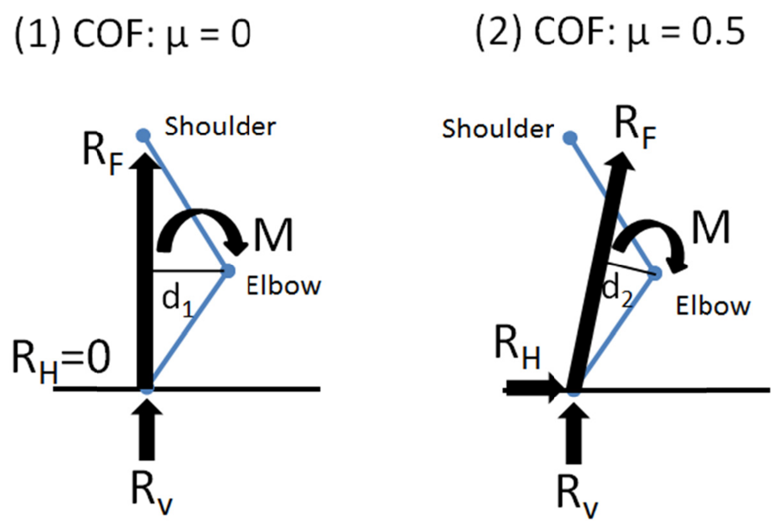


**Figure 1.3** Comparison example two cases (1) COF : $\mu = 0$ (2) COF : $\mu > 0.5$ in terms of the applied moments on the elbow joint.

In terms of the effect of the landing surface, if the coefficient of friction (COF) is less than 0.5, it represents a slippery surface such as ice. On the icy surface, the result force ($R_F$) on the wrist results only from the normal force ($R_V$); multiplying this force by its lever arm about the elbow joint yields a larger elbow flexion moment (M) (Figure 1.3, left) than on the non-slip surface (Figure 1.3, right). So arresting a fall on a non-slip surface may require less elbow extensor strength than on the slippery surface, but greater shoulder muscle strength.

Surface padding can reduce peak forces applied to the hands by 35% (Robinovitch and Chiu, 1998). The importance of soft tissue energy absorption on the potential to fracture the wrist was demonstrated by Nikolić et al. (1975). These results indicate that a large portion of energy was dissipated by the soft tissues of the hand closest to the site of impact. Robinovitch et al. (1995) documented the same effect at the hip. Additional experimental studies have found that peak forces can be substantially modified even when the surface and soft tissue conditions remain the same. When five healthy young males were released from a forward lean so that they fell 40 cm onto a padded surface, peak forces on each hand varied from 0.7 to 1.5-times body weight (Figure 1.1). Upon receiving verbal instruction and a visual demonstrations on how to minimize the impact force in such falls, they were able to significantly reduce the peak hand impact force by an average of 27% within only four trials, compared to the impact forces measured in a self-selected ‘‘natural’’ landing (DeGoede and Ashton-Miller, 2002). So, both the nature of the impact surface and the strategy used to arrest the fall can affect the magnitude of the impulsive loading on the hands in a fall.

### 1.2.2 Experimental Study of Forward Falls

The biomechanical factors affecting the magnitude of impact forces on the distal forearm in forward falls have started to be identified. DeGoede and Ashton-Miller (2002) studied the age differences in rapid arm movement prior to arresting an impending pendulum. They showed that age, gender and perceived threat significantly affected movement times. Others (for example, Chiu and Robinovitch, 1998) studied forward falls from a wrist height up to 5 cm above ground and they concluded that fall heights greater than 0.6 m carry significant risk of wrist fracture. DeGoede and Ashton-Miller (2002) showed that young males can volitionally reduce the wrist impact forces by pre-impact adjustment of the elbow angle and wrist impact speed in forward falls from 1 m shoulder height. However, the wrist impact force reduction was obtained by using a more flexed elbow angle upon impact, and this could require substantial arm strength in order to the elbow from buckling.

Robinovitch et al. (2005) showed that healthy older women, although slower than young women in initiating hand movements, were generally able to move the hands quickly enough into an appropriate position for breaking a forward fall. Recently, Sran et al. (2010) showed that the energy-absorbing capacity of the upper extremities in older women was nearly half that of young women during a push-up task which simulated a fall arrest. This might relate to the increase in the increasing prevalence of fall-related hip fracture with age (Cummings et al., 1989; Jaglal et al., 2005).

### 1.2.3 Computer Simulation Studies of Forward Falls

A forward fall simulation study (DeGoede and Ashton-Miller, 2003) predicted that the age-related reduction in upper extremity extensor muscle strength can significantly reduce the ability of older women to safely arrest a fall. For example, this reduction could mean that a slightly-flexed elbow might buckle, thereby increasing the risk of a head/torso impact. On the other hand, if older women straighten the elbow in order to reduce this risk, then they risk fracturing the distal forearm because of the increased stiffness of that configuration of the upper extremity skeleton. To address this conundrum, we shall explore the factors that critically affect the ability to arrest a fall on the hands.

Construction of upper extremity models for analyses of the biomechanics of the impact phase of fall arrests requires, among other data, data on body segment anthropometry, as well as joint stiffness and damping properties. Anthropometric data are readily available (for example, Kroemer and Kroemer, 1997), but Chambers et al. (2010) concluded that age, obesity and gender had a significant impact on segment mass, center of mass and radius of gyration in older adults. Therefore, we need to consider age, obesity and gender when thinking about the anthropometrics of an aging population. In addition, little is known about the stiffness and damping properties of the muscular soft tissues acting about the upper extremity joints during impact or pre- or post-impact. So measurements of arm rotational stiffness and damping parameters under impulsive end-loading will be made as a part of this dissertation.

### 1.2.4 Study of Lateral Falls

Of the few biomechanical studies of lateral falls, both theoretical (van den Kroonenberg at al., 1995) and experimental (Sabick et al., 1999; Robinovitch et al., 2003; Groen et al., 2007) investigations have shown that active responses during a lateral fall can alter the resulting impact forces. Lo and Ashton-Miller (2008) simulated several different lateral fall strategies (for example, free falls, broomstick falls with/without arm, hip lateral flexion falls with/without arms). They found that (a) the whole body kinetic energy at hip contact greatly depends upon whether hip was the first impact site or not, (b) the orientation of the pelvis at the instant of impact also played a significantly role in determining the impact severity to the hip during, and (c) the importance of the decision-making time in determining impact injury risk, which is in agreement with previous epidemiology studies (for example, Nevitt et al., 1989).

Some studies (for example, Choi et al., 2010; Laing and Robinovitch, 2008) have tested the effectiveness of soft hip protectors for the elderly to examine how a soft shell hip protector affects the magnitude and distribution of force to the hip during simulated falls, and how the protective effect depends on the fall direction and the amount of soft tissue padding over the hip. More studies have examined forward than lateral falls, even though it is the lateral falls that cause hip fractures (Grisso et al., 1991) and hip fractures are more than five times more costly than wrist fractures (Gabriel et al., 2002; Ray et al., 1997). Cleary, more research is needed on the mechanics of lateral falls.

## 1.3 Conceptual Models for this Dissertation

### 1.3.1. Study of Elbow Muscles

A number of studies have examined the mechanics of how forward falls are arrested by males (for example, DeGoede and Ashton-Miller, 2002; Lo et al., 2003) or females (for examples, Sran et al., 2010; Robinovitch et al., 2005). There are, however, no reports of how both gender and age may affect the mechanics of the upper extremities during forward fall arrests. Before we address the working hypotheses in this dissertation, let us consider the following conceptual model diagram (Figure 1.4).

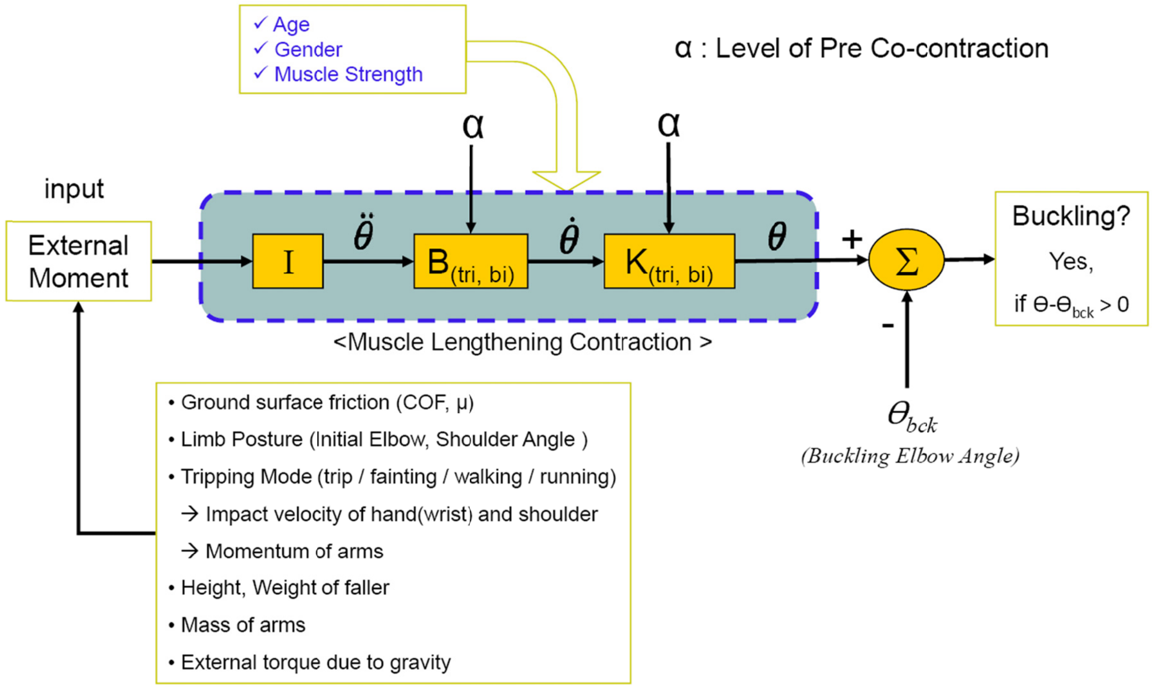


**Figure 1.4** Conceptual model diagram showing information flow among the components of the muscle joint control system when arresting a fall. See text for explanation of symbols.

When one perceives the onset of a fall and the two hands strike on the ground, the ground reaction force causes an external flexion moment on the upper extremities. The limb dynamics including velocities and positions of limbs and joints are determined with this input and the inertia (I) of limbs, as well as by age and gender, the level of pre-co-contraction (α), and muscle strength (Figure 1.4). We intend to examine how these factors affect the rotational stiffness (K) and damping (B) resistance to impulsive elbow flexion loading. In addition, the external moment could be affected by several other factors: the ground surface friction (μ), the initial configuration of limbs, the mode of tripping, the mass of arms, the height/weight of faller, and the external torque due to gravity (Figure 1.4). If, under the impulsive load, the output elbow angle (Θ) exceeds the nominal elbow buckling angle ($\Theta_{bck}$), then we can say the arms are buckling.

In order to avoid elbow buckling during a fall, we need to measure the properties of elbow joint rotational stiffness (K) and damping (B) which play an important role to deciding how the elbow flexion angle (Θ) changes with time during the fall arrest.

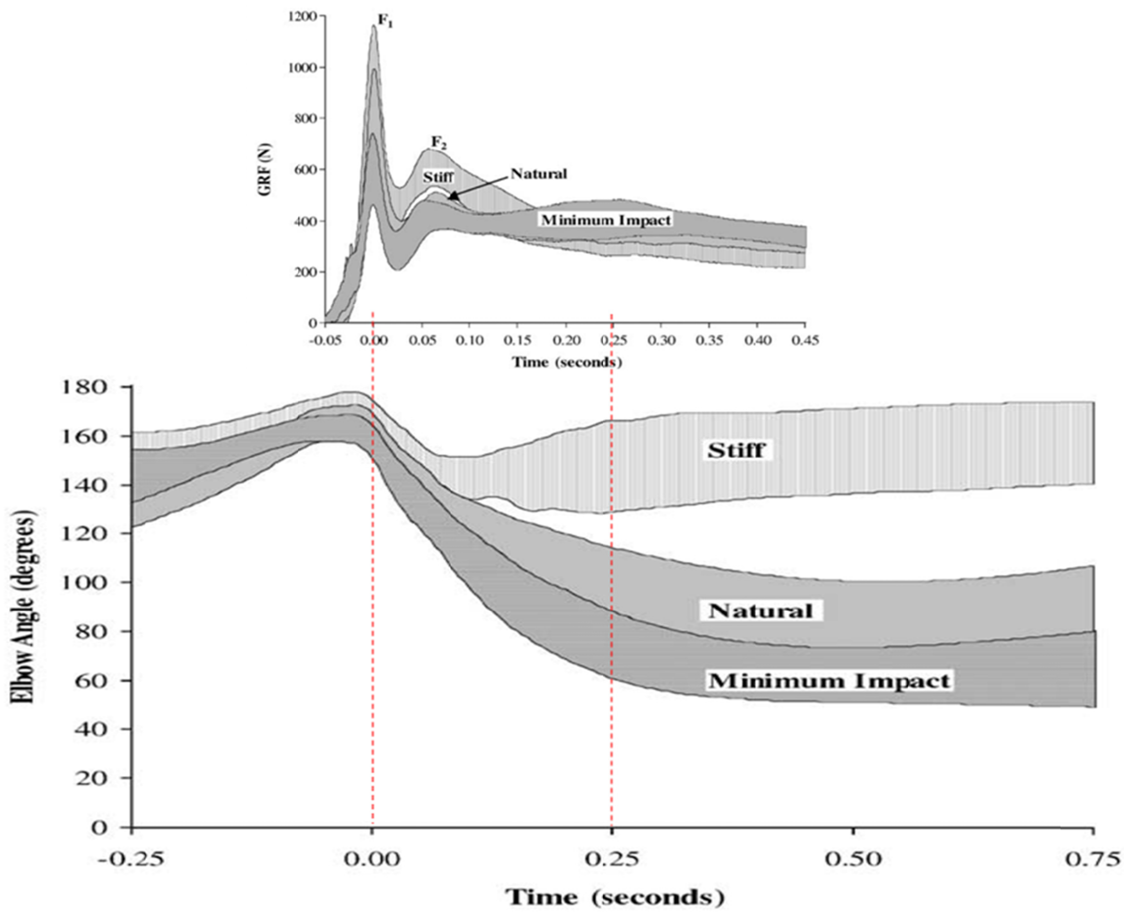


**Figure 1.5** Overlap result of time history of the impact force (N) on one wrist and the elbow angle deflexion (degree) on the same time span (y-axis) (DeGoede and Ashton-Miller, 2002).

If the two experimental graphs are overlapped on the same time course (Figure 1.5), we notice that the total time from the impact to arrest a fall is less than 700ms in a forward falling. Let us consider what happens at the elbow joint. During this time, the subject pre-contracted upper body muscles including the triceps brachii, biceps brachii and shoulder muscles. The triceps brachii plays an important role in this contractile mechanism as an agonist because the muscle-tendon complex of the triceps brachii does undergo a lengthening contraction during this phase and the biceps brachii acts as an antagonist during the lengthening contractions. Since pre-activation is induced by primarily triceps brachii (agonist) during lengthening contraction, the effect of antagonist (biceps brachii) is likely small (Figure 1.3) since muscle cannot push. Therefore, we need to focus on the properties of the pre-activated elbow agonist muscles (for examples, triceps brachii ($K_{tri}$, $B_{tri}$), and the anterior deltoids at the shoulder joints. The goal would be to determine the rotational stiffness (K) and damping (B) properties of these co-activated muscles under the impulsive loading associated with arresting a fall.

### 1.3.2 Study of Shoulder Muscles

In terms of the muscle cocontraction of upper extremity during impact, the shoulder muscle properties are another important factor that likely determine the behavior of upper extremity under impulsive load. As mentioned Section 1.3.1 the elbow triceps brachii are the main muscles that determine the stiffness and damping properties of the elbow, while the anterior deltoid and pectoralis major play a similar role for the shoulder when arresting a forward fall.

Due to the complexity of the shoulder structures, it is difficult to quantify the kinematics of the bones comprising the shoulder during impact without stereoradiography. We therefore developed a custom drop-weight apparatus (Chapter 4) to be able to study the movement of shoulder in two of its three planes: shoulder flexion represented in the sagittal plane and shoulder abduction represented in the frontal plane.

### 1.3.3 Gender and Pre-contraction Level Effect of Shoulder and Elbow Responses

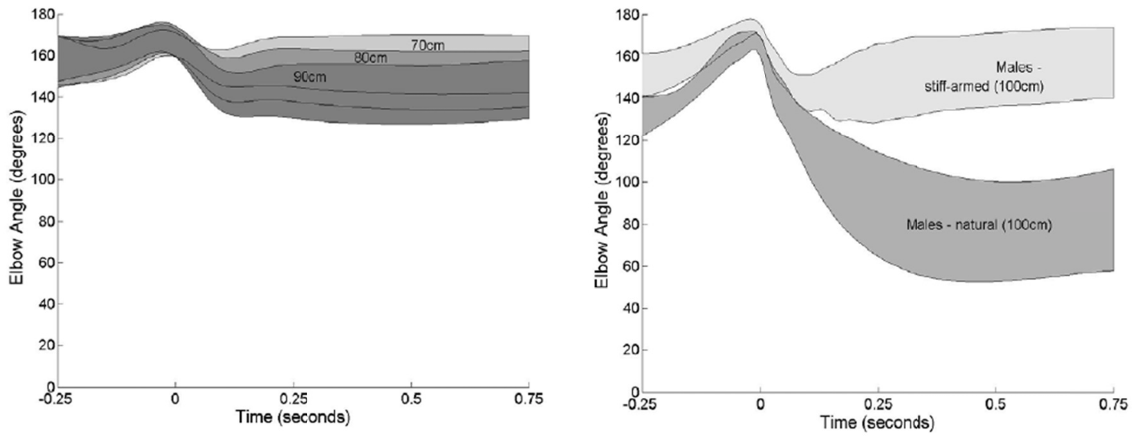


**Figure 1.6** Time history of the elbow angle deflexion (degree) on young females (left: Case et al., 2005) and on young males (right: DeGoede and Ashton-Miller, 2002).

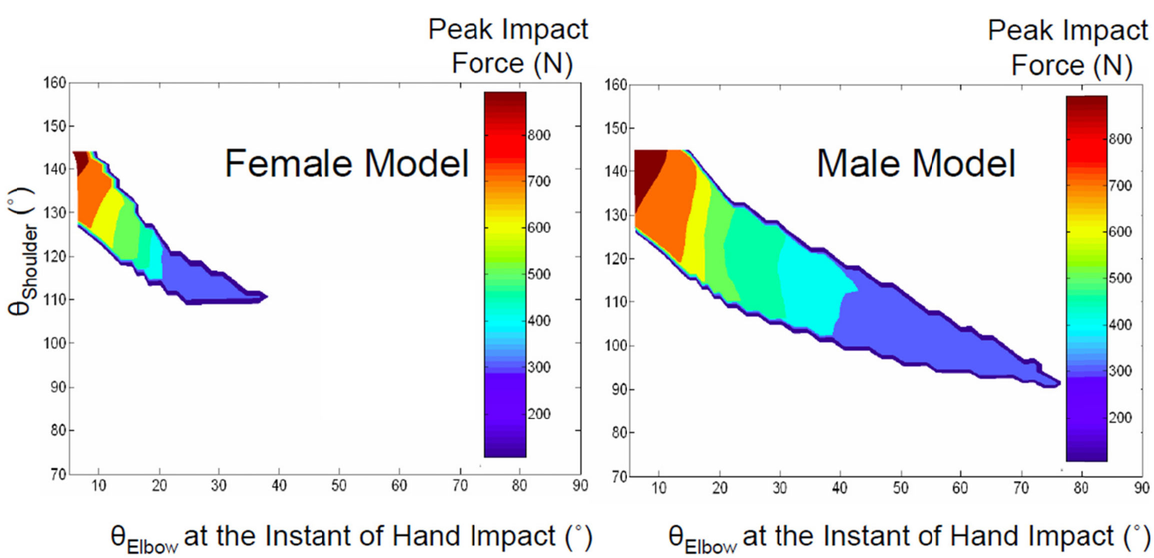


**Figure 1.7** The peak impact force (N) on the wrist is shown in color along with the elbow and shoulder deflexion angle in young females (left: Case et al., 2005) and young males (right: DeGoede and Ashton-Miller, 2002).

Case (Case, et al., 2005) used the DeGoede forward fall paradigm (DeGoede and Ashton-Miller, 2002) to investigate young females' natural responses to forward falls. They found there was gender difference in the elbow angle changes after the wrist impact (Figures 1.6 and 1.7).

In Figure 1.7, we can see that the predicted peak impact force on the wrist depended on pre- and post- cocontraction muscle level of upper extremity because the configuration was the result of three different contraction levels (the 'natural', 'stiff-arm' and 'minimum-impact' trials) in young males. Therefore there was the relationship between the peak impact force and joint deflexion angles which exhibited a significant gender difference. It is notable that young females exhibited a smaller range of angular deflexion in the elbow and shoulder during impact.

### 1.3.4 Non-linearity of Muscle Properties

When an upper extremity is used to arrest a fall to the ground, the impulsive end-load at the wrist can reach one body-weight (1*BW) or more (DeGoede and Ashton-Miller 2002, Tan et al. 2006, Lo and Ashton-Miller 2008). If the upper extremity were to give way or buckle under such a load then there is an increased risk of the head striking the ground. In fact head impact is indeed a phenomenon known to be associated with falls in the elderly (Kannus et al. 2007), and it increases markedly with advancing age (Jacobsson et al. 2007).

Part of the reason for the increased risk of head injury with age may be the thirty percent decrement in arm extensor muscle strength that occurs between 50 and 80 years (Metter et al. 1997). This would increase the risk of the extremity collapsing under a large impulsive end-load because striated muscle tensile stiffness is known to be proportional to the force developed by the muscle (i.e., Blanpied & Smidt 1993, Ettema and Huijing 1994). So reduced extensor muscle stiffness, and possibly reduced damping, can be expected to translate into a reduced upper extremity buckling load.

We are not aware of any estimates of how muscle elastic or viscous properties affect the behavior of the upper extremity under an impulsive end-loaded. Active striated muscle is known to exhibit non-linear stretch behavior (Grover et al., 2007 and Malamud et al., 1996). Therefore in this dissertation we need to explore the effect of non-linear stretch behavior of muscle on arm behavior under impulsive end loading as determined by the relationship between muscle-tendon unit contractile force and its tensile stiffness and damping coefficients.

### 1.3.5 Safe vs. Unsafe Conditions and the Buckling Elbow Angle

We can define an arm to have buckled if the elbow deflexion angle ($\Theta$) exceeds the nominal elbow buckling angle ($\Theta_{bck}$) which is unsafe under an impulsive end-load. And we can say the peak reaction force at the wrist is the buckling load which would possibly cause that the head hits to the ground. We will focus on how the upper extremities behave after the instant of impact and what biomechanical factors affect whether the arm is buckled or not in this dissertation.

For example, the man of body height 190 cm falling forward, with the hands outstretched, onto the ground (Figure 1.8). Let us say that he hits the ground with an almost straight elbow angle, $\Theta = 162°$ and then the shoulder height, h is 652 mm (Figure 1.8 (A)). If his arms flex until the elbow angle, $\Theta^* = 52°$ which is the elbow deflexion angle ΔΘ = 110 °, his shoulder height is reduced to h* = 290 mm or 55% of the original shoulder height, h (Figure 1.8 (B)). If the shoulder height becomes any lower than h*, the man is unsafe because his head will strike the ground if he allows his neck to flex. Now we call these boundaries as the critical conditions such as the critical angle, Θ* or the critical shoulder height, h*.

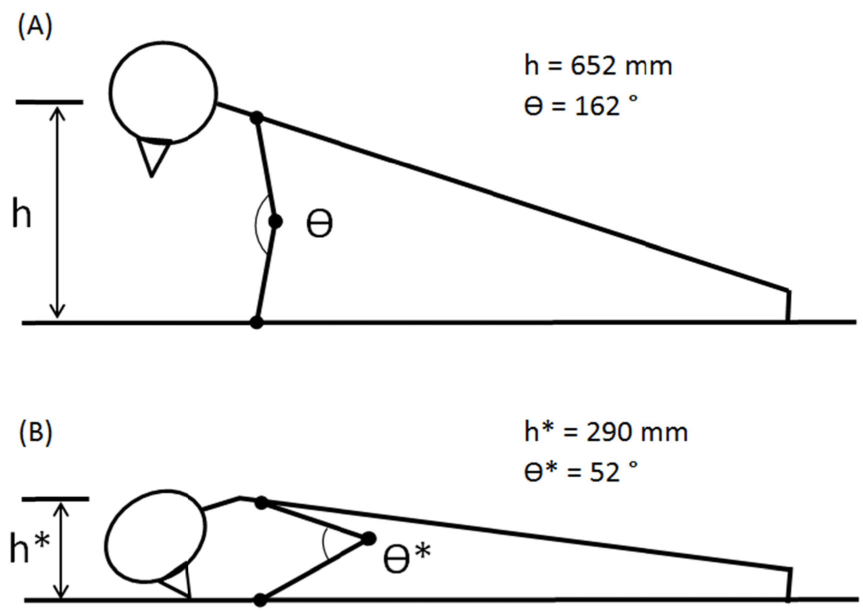


**Figure 1.8** Conceptual diagram for arresting a fall while the hand hitting the ground. (A) shows the impact moment with the straight neck posture (the shoulder height, h and the elbow angle, Θ) (B) represents the critical condition which is considered as arm buckled (or head hit) with the fully flexed neck posture, the critical height, h* and the critical elbow angle, Θ*. These measurements were taken from a man with a stature of 190 cm.

The typical ground reaction force at wrist while arresting a fall to the ground (DeGoede et al, 2003) is shown in Figure 1.9 (a). The elbow angle is decreased from Θ to the critical elbow angle, Θ* which we consider an unsafe region in Figure 1.9 (b) in other words, the arms have buckled. The elbow moment (c) is a function of the ground reaction force, the limb length, the joint angle, the muscle properties about the joint which are represented as stiffness (d) and damping resistance (e) in Figure 1.9. These muscle properties would be set by neural system as a pre-contraction before impact. Grover et al., (2007) found that the striated muscle stiffness exhibits a bi-linear characteristic when actively stimulated whole muscle is suddenly stretched. From their graph (Figure 1.10 in Grover et al. 2007), we conclude that the break-point in the stiffness was at approximately 14% of the full range of motion (i.e., muscle length or joint angle). This characteristic will be studied in Chapter 3 using three different simulation models. We will explore the effect of non-linearity on the rotational stiffness and the rotational damping of each joint on the behavior of upper extremity under an impulsive load; Model I is given a linear stiffness and a linear damping coefficient; Model II has a non-linear stiffness and a non-linear damping; and Model III has a bi-linear stiffness and a linear damping coefficient.

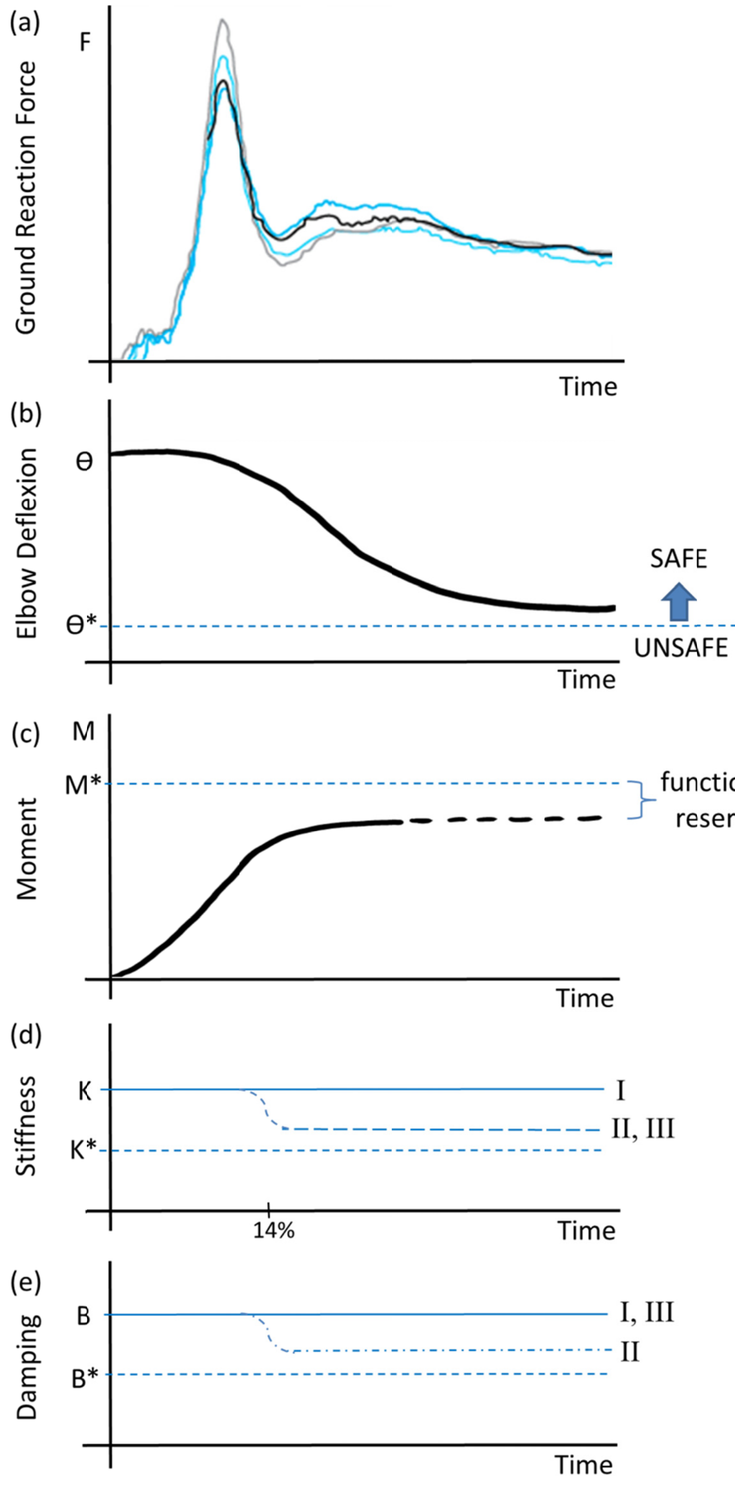


**Figure 1.9** Representative biomechanical factors to arrest a fall in Figure 1.8: (a) the ground reaction force at the wrist, (b) the elbow deflexion angle, (c) the elbow moment, (d) the elbow muscle stiffness, and (e) the elbow muscle damping coefficient.

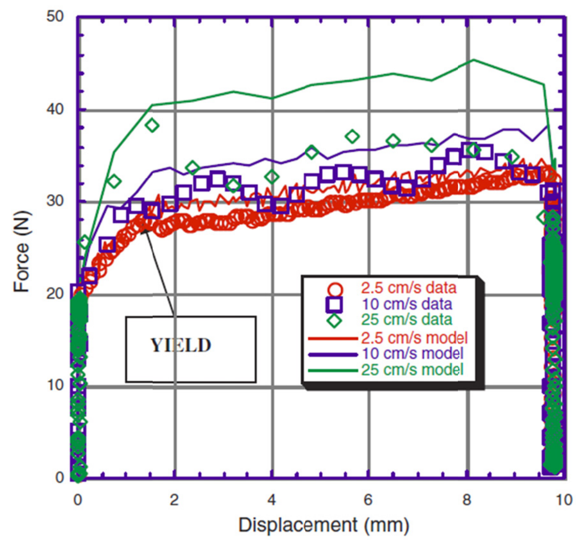


**Figure 1.10** Force vs. displacement data for tetanized muscle at constant velocity exhibited yielding and the dynamic systems simulation of the rate dependent force-displacement behavior in rabbit tibialis anterior muscle (Grover et al., 2007).

### 1.3.6 Study of Proximal Force Propagation Along the Upper Extremity

The upper extremities are typically the first line of defense to protect the head and torso while bracing for a frontal car crash (Frampton et al., 1997) or as well arresting a fall to the ground (DeGoede et al, 2003). Bracing for a car crash or falling onto an outstretched arm can be associated with wrist, elbow and/or shoulder injury (Loder and Mathew, 1988; Skyhar et al., 1990). Therefore using a slightly flexed limb would logically be used to prevent from buckling under an impulsive load applied to the hand (DeGoede et al., 2002; DeGoede and Ashton-Miller 2002; Lee and Ashton-Miller 2011) especially if the arm muscles are precontracted to brace the arm for impact (DeGoede et al, 2003; Lo and Ashton-Miller, 2008; Lo et al., 2003; Troy and Grabiner, 2007).

If the muscle pre-contraction is not used, then one has to rely on neuromuscular reflexes to increase muscular resistance to arm buckling. Are the muscle reflexes sufficiently rapid that they can effectively increase muscle stiffness before the muscle is forcibly stretched by limb flexion under gravitational and inertial loading? And to what extent are the force transmitted proximally along the limb and what exactly time to flex when an arm is end-loaded? These questions will be addressed in this dissertation because they may all affect the impulse response of the end-loaded upper extremity.

### 1.4 Working Hypotheses and Primary Goals

In this dissertation, we explored the working hypotheses that arm protraction muscle strength and pre-contraction state largely determines the buckling behavior of the end-loaded arm, and that gender and advancing age both adversely affect this behavior.

In Chapter 2, the primary hypotheses were tested that neither (1) gender, (2) level of cocontraction, nor (3) initial elbow angle affect the rotational stiffness or damping coefficients of the extensor muscles acting about the elbow in healthy young adults.

In Chapter 3, we explored the effect of (1) age and gender, (2) non-linear muscle stretch responses and (2) elbow angle-dependent triceps moment arm on the load-displacement behavior of the upper extremity axially end-loaded by an impulsive load using three computer simulation models.

The goal of Chapter 4 was to determine how impulsive end loading affects the behavior of the young and old, female and male adults arm given low to moderate levels of muscle co-activation, and how that behavior depends on muscle strength. We developed a lever arm drop-weight apparatus to test the effect of age, gender, level of cocontraction on the upper extremity impulse response behavior.

In Chapter 5, we used this apparatus to test the primary hypotheses that neither gender, age, nor level of pre-cocontraction affect the time it takes an impulsive force to propagate proximally along the upper extremity in healthy adults based on Chapter 4. The secondary hypothesis was that this propagation time is always shorter than the latency of the triceps EMG response to elbow flexion caused by impulsive loading. To help interpret the results, a forward dynamics model was used to explore how the magnitude of hand preload affects impulse propagation times along the upper extremity.

In Chapter 6 (General Discussion) we state what is new about the findings of this dissertation and pull together the findings from Chapters 2 through Chapter 5 so we can interpret them in terms of what is known in the literature. We also discuss the strengths and weakness of the overall approach and possible avenues for further research in this Chapter. In Chapter 7 (Conclusions) we briefly summarize the main findings from each chapter as well as from combinations of chapters.

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# CHAPTER 2

## The Effects of Gender, Level of Co-Contraction and Initial Angle On Elbow Extensor Muscle Stiffness and Damping Under a Step Increase in Elbow Flexion Moment

### 2.1 Abstract

Flexion-buckling of an arm under the large ground reaction loads associated with arresting a fall to the ground increases the risk for head and thorax injuries. Yet the factors that determine the arm buckling load remain poorly understood. We tested the hypothesis in 18 healthy young adults that neither gender, triceps co-contraction level (i.e., 25, 50 or 75% MVC) nor elbow angle would affect the rotational stiffness and damping resistance to step changes in elbow flexion loading. Data on the step response were gathered using optoelectronic markers (150 Hz) and myoelectric activity measurements (2 kHz), and an inverse dynamics analysis was used to estimate elbow extensor stiffness and damping coefficients. A repeated-measures analysis of variance showed that gender ($p = 0.032$), elbow flexion angle and co-contraction level (both $p < 0.001$) affected stiffness, but only the latter affected the damping coefficient ($p = 0.035$). At 25 degrees of initial elbow flexion angle and maximum co-contraction, female

stiffness and damping coefficients were 18 and 30% less, respectively, than male values after normalization by body height and weight. We conclude that the maximum extensor rotational stiffness and damping at the elbow is lower in women than in men of the same body size, and varies with triceps co-contraction level and initial elbow angle.

## 2.2 Introduction

Falls are a leading cause of injury in the population (CDC Injury Research Agenda 2009-2018). One or both upper extremities is often used to protect the head and thorax in a fall. However, the loads imposed on the upper extremities can be substantial, with the step end-load at each wrist reaching one body-weight (1*BW) or more in a fall to the ground (Tan et al., 2006; Lo et al., 2003; DeGoede and Ashton-Miller, 2002). If the upper extremity were to flexion buckle ("give way") at the elbow under such loading then there is an increased risk of the head striking the ground. Indeed, the elderly are particularly prone to fall-related head (Kannus et al., 2007a; Kannus et al., 1999) and cervical spine injuries (Kannus et al., 2007b). A current knowledge gap includes the factors that determine the threshold load required to flexion buckle the elbow of an end-loaded human arm.

Direct measurements of the magnitude of the end-load required to initiate arm flexion buckling are contraindicated because of the risk for stretch-related injury of the elbow extensor muscles, their tendons, and/or their entheses. But the risk for stretch-related injury of active striated muscle is known to increase with the product of tensile force and elongation strain in striated muscle (Brooks et al., 1995). So, an indirect approach to estimating the threshold for flexion-buckling seems warranted. A logical first step, the goal of this paper, is to measure the elastic and viscous resistance of the actively contracting elbow extensor muscles to small sudden stretches.

From first principles the flexion buckling load of the end-loaded arm will depend upon the rotational elastic and damping resistance of the elbow extensor muscles to forced flexion, the initial elbow angle and initial arm extensor co-contraction level (DeGoede and Ashton-Miller, 2002), their moment arm about the elbow, the segmental lengths and inertias of the forearm and upper arm, and the magnitude of the elbow flexion moment caused by the direction of the end-load. The voluntary 'co-contraction' of the muscles acting about a joint is associated with an increase in muscle recruitment of both agonist and antagonist muscle (Kearney and Hunter 1990). Such co-contraction is known to increase rotational stiffness at the elbow and knee (Granata et al., 2004; Wojtys et al., 2003; Lacquaniti et al., 1982). For various initial elbow flexion angles and degrees of elbow extensor muscle co-contraction, one can apply step elbow flexion moments to measure the kinematic response in elbow flexion. A planar, lumped parameter, model can then be used to estimate the rotational stiffness and damping resistance of the elbow extensor muscles to resist step elbow flexion, and show how these depend upon muscle co-contraction level and initial elbow angle.

It is known that there is a significant gender difference in arm extensor strength in healthy adults (Chaffin et al., 2006; Stobbe 1982). Females have a smaller arm muscle mass than males (Gallagher et al., 1997). So, one can hypothesize that the maximal elastic and viscous muscular resistance to forced elbow flexion should also be less in the female, since the resistance to stretch of a muscle depends on the number of cross-bridges in the strongly bound state: at the same contraction intensity, smaller diameter muscles should have fewer strongly bound cross-bridges than larger muscles with the same architecture. For both genders, the non-dominant arm is the most likely to buckle because its elbow extensor strength (and stiffness) is typically at least 5% less than the dominant arm (Askew et al., 1987).

The goal of this paper, therefore, was to test the three hypotheses that neither gender, level of triceps co-contraction nor initial elbow angle affect the rotational stiffness or damping coefficients of the extensor muscles acting about the non-dominant elbow in healthy young adults.

## 2.3 Methods

Nine healthy young males of mean (SD) age of 25.1 (4.4) years and nine healthy young females of 20.3 (2.4) years gave written informed consent to participate in the study, which was approved by the institutional review board. Mean height and mass for the males were 1.75 (0.074) m and 76.2 (17.3) kg, respectively, and for the females were 1.61 (0.068) m and 57.3 (8.35) kg, respectively. Subjects were screened by telephone to exclude any chronic illnesses or upper extremity fractures or sprains within the previous year.

The subject was seated with the left upper arm supported by a pad and with the elbow at either $10^{o}$ or $25^{o}$ flexion in the horizontal plane (where $0^{o}$ is full elbow extension) (Figure 1). The horizontal plane was used to obviate gravitational effects on the measurements. A strain-gauged load cell (1,779 N capacity) was attached to the wrist via a wrist cuff and then connected via aircraft cable to a weight pan weighing 1.4 kg. A solenoid-actuated crossbow release mechanism was used to release a cable to drop a weight of 5.2 kg onto the weight pan in order to cause the step elbow flexion moment via the wrist cuff.

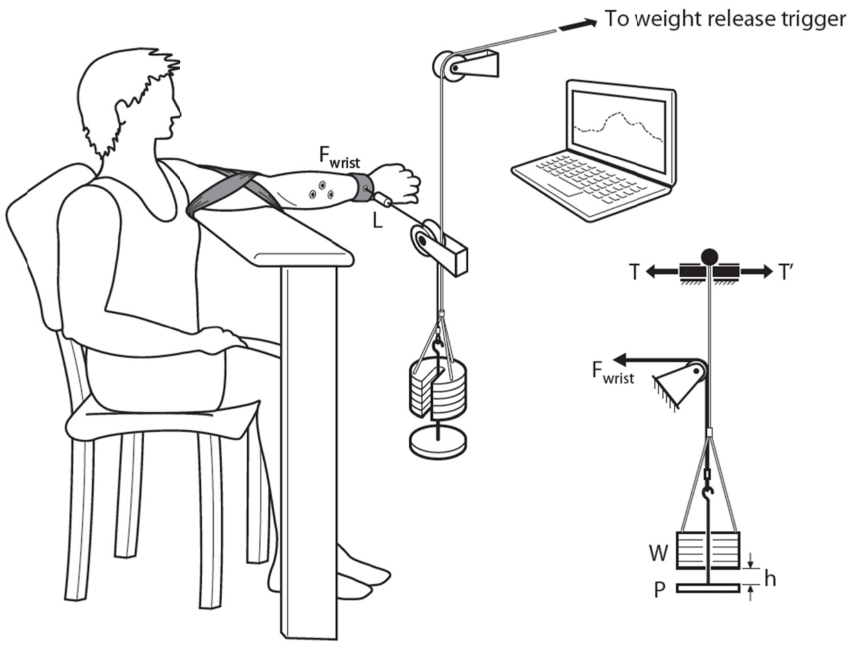


**Figure 2.1** Experimental set-up showing a subject maintaining 10o elbow flexion while resisting the baseline elbow flexion moment with a specified co-contraction of their elbow muscles using visual biofeedback of their triceps EMG. In the next instant the weight, W, will be dropped suddenly using the trigger force, T, so that it lands on the weight pan, P, to apply a step elbow flexion moment via the wrist force, Fwrist , applied through load cell, L.

Bipolar surface electromyographic electrodes, spaced 2 cm apart, were applied to the short head of the biceps brachii and the lateral and long heads of the triceps brachii. The kinematic response of the forearm was measured at 150 Hz using marker triads and an Optotrak 3020 (Northern Digital, Inc. Waterloo, Canada) optoelectronic camera system. The impact force was measured by the load cell at 2 kHz. The kinematic and force data were digitally low-pass filtered (MATLAB 2007a, The MathWorks) with a Butterworth filter and cutoff frequencies of 10 Hz and 8 Hz, respectively. Integrated surface electromyography (EMG) data were collected at 2 kHz using an amplifier system (Myosystem 2000, Noraxon Inc., Scottsdale, Arizona, USA).

In order to find the maximum possible triceps muscle root-mean-square (RMS) EMG activity for normalization purposes, subjects performed three different tests with the elbow at $25^{o}$ flexion. In the first test, subjects stood with their back to a wall and slowly exerted maximum bilateral arm protraction strength by pushing forward bimanually against a horizontal, 2.5 cm-diameter, metal bar with hands at shoulder width and height; a fixed length of aircraft cable attached each end of the bar to the wall via frictionless pulleys and a 1,799 N load cell. Maximum protraction strength was recorded. In the second test, the subjects exerted unilateral maximum protraction strength against resistance provided by an examiner. In the third test, the subject sat (Figure. 2.1) and maximally co-contracted their arm and upper body muscles. Each subject performed three 5-second trials separated by 2 minutes rest intervals. Subsequent EMG data were normalized by the maximum RMS EMG value found in any of these tests.

To determine elbow extensor muscle rotational stiffness and damping parameters, each subject was seated with the left upper arm supported and strapped to a special table to prevent it from moving (Fig. 1). The forearm was free to move in the horizontal (gravity-free) plane. A load cell strapped to the subject's wrist was connected via 3.2 mm-diameter aircraft cable and frictionless pulleys to a 1.4 kg weight pan. The subject viewed RMS EMG biofeedback from the lateral triceps brachii and was instructed to keep his/her muscle activation at a specified level prior to and throughout the entire impact response trial. Test instructions were “not to intervene” when the arm was perturbed so the stiffness and damping associated with the triceps pre-contraction could be measured. Once (s)he had achieved the specified EMG level, a weight of 5.2 kg was released after a random delay to fall onto the weight pan, thereby causing a step increase in elbow flexion angle. Practice trials were performed to familiarize the subjects. Then, four trials each were performed in each of six conditions with the order of presentation randomized: at 25, 50, and 75% MVC with the subject’s initial elbow flexion angle at $10^{o}$ (where $0^{o}$ is full extension), and at 25, 50 and 75% MVC with the subject’s initial elbow flexion angle at $25^{o}$. Extra trials were sometimes indicated if the required EMG levels were not met satisfactorily or if there was a technical problem with the data acquisition.

**Data Analyses**

A second-order, planar, rotational spring-damper model was used to approximate the dynamic properties of the elbow joint: $T = I\ddot{\theta} + B\dot{\theta} + K\theta$, where T is the applied flexion torque, $\theta$ is the angular displacement of the elbow joint, $\dot{\theta}$ is the angular velocity of the elbow joint, $\ddot{\theta}$ is the angular acceleration of the elbow joint, I is the calculated moment of inertia of the limb (forearm + hand), B is the damping coefficient, and K is the stiffness coefficient. The moment of inertia was calculated by measuring the length of the limb and knowing the location of the center of mass of hand and forearm link (Chaffin et al., 2006).

A deterministic global optimization algorithm (‘gclSolve.m’ in Matlab$^{TM}$) was used to determine K and B using the experimental applied flexion moment (torque), the limb inertia, joint angle, angular velocity, and angular acceleration. This was done by minimizing the square of the paired differences between corresponding points on the measured and calculated elbow torque-angle curves (see Fig. 2). The values of K and B were normalized by subject body weight and height to reduce the effect of body size as a confounder in inter-subject comparisons and make the data more transferrable to other subjects.

Descriptive statistics were calculated for K and B. A repeated measures analysis of variance was used to test for effects of gender, three different muscle co-contraction levels, and two different initial elbow joint angles on K and B using SAS 9.1 software. A p-value of less than 0.05 was considered statistically significant for the three main effects (primary hypotheses). A Bonferoni correction was used for secondary hypothesis testing (interactions).

## 2.4 RESULTS

Sample elbow flexion moment-angle relationships for a single step elbow response trial in a male and a female subject, and the corresponding model-predicted response, are shown in Figure 2.2. The hysteresis is evidence of the damping behavior of the elbow extensor musculature. Table 2.1 shows the mean normalized values of elbow stiffness and damping found during each of the six testing conditions (three muscle co-contraction levels and two different initial elbow angles) for each gender. Increasing the initial flexion angle tends to increase stiffness by 15-48%, but not the damping coefficient, in both male and female subjects. The males mostly had higher stiffness and damping values than the females.

The repeated measures ANOVA (Table 2.2) led us to reject all three hypotheses: gender significantly affected normalized elbow stiffness; stiffness increased significantly with the muscle co-contraction level and with the elbow angle. Damping behavior was not significantly altered by gender or elbow angle, but was altered by co-contraction level.

Sample moment-angle relationships show large differences at three different triceps co-contraction EMG levels (Figure 2.3). The means of normalized stiffness of each condition were 0.791 Nm/rad/kg/m in 71-90% MVC, 0.534 Nm/rad/kg/m in 51-70% MVC, and 0.267 Nm/rad/kg/m in 31-50% MVC. The means of normalized damping are 0.020 Nms/rad/kg/m, 0.011 Nms/rad/kg/m and 0.009 Nms/rad/kg/m, respectively. The stiffness and damping coefficient values increased with the higher EMG co-contraction levels.

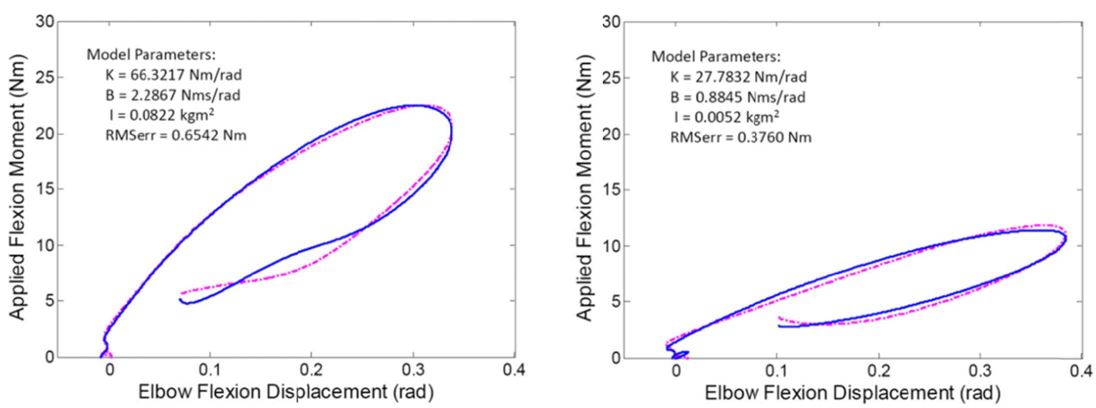


**Figure 2.2** Sample measured (broken line) and fitted (solid line) elbow flexion moment vs. elbow flexion displacement relationships from one male (left panel, subject BCAB) and one female (right panel, subject BGAA). In this and the following figure data were taken starting from an initial angle of 25 degree elbow flexion with triceps muscle co-contracted between 51-70%. Model parameters are also given.

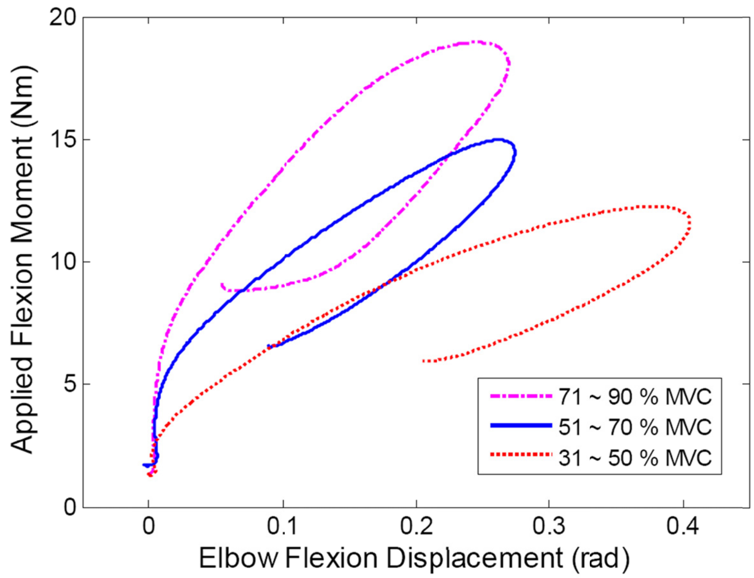


**Figure 2.3** Sample plots showing the effect of three different levels of triceps muscle co-contraction on the applied flexion moment vs. elbow angular displacement starting from an initial angle of 10 degrees flexion. The trials are from male subject BCAI.

**Table 2.1** Means (SD) normalized elbow rotational stiffness and damping coefficients by gender and level of triceps co-contraction.

| | | Normalized Stiffness (Nm/rad/kg/m) | | | | Normalized Damping (Nms/rad/kg/m) | | | |
|---|---|---|---|---|---|---|---|---|---|
| Normalized Co-Contraction EMG Level | Elbow Angle (degree) | **Males** mean | (SD) | **Females** mean | (SD) | **Males** mean | (SD) | **Females** mean | (SD) |
| 31-50 % MVC | 10 deg | 0.370 | ( 0.160 ) | 0.268 | ( 0.122 ) | 0.019 | ( 0.016 ) | 0.012 | ( 0.005 ) |
| 51-70 % MVC | 10 deg | 0.419 | ( 0.166 ) | 0.435 | ( 0.218 ) | 0.026 | ( 0.018 ) | 0.013 | ( 0.005 ) |
| 71-90 % MVC | 10 deg | 0.596 | ( 0.219 ) | 0.459 | ( 0.292 ) | 0.022 | ( 0.015 ) | 0.017 | ( 0.005 ) |
| 31-50 % MVC | 25 deg | 0.357 | ( 0.100 ) | 0.427 | ( 0.192) | 0.017 | ( 0.013 ) | 0.007 | ( 0.004 ) |
| 51-70 % MVC | 25 deg | 0.622 | ( 0.177 ) | 0.520 | ( 0.189 ) | 0.018 | ( 0.015 ) | 0.012 | ( 0.004 ) |
| 71-90 % MVC | 25 deg | 0.688 | ( 0.208 ) | 0.564 | ( 0.274 ) | 0.023 | ( 0.018 ) | 0.016 | ( 0.006 ) |

**Table 2.2** ANOVA tables for the main effect and interaction for normalized elbow stiffness and damping coefficients. (*p < 0.05)

| | *Stiffness* | | *Damping* | |
|---|---|---|---|---|
| **Main Effect** | F | P | F | P |
| Gender | 5.58 | 0.032* | 2.66 | 0.124 |
| Co-contraction | 26.44 | <.001* | 8.28 | 0.004* |
| Elbow Angle | 30.56 | <.001* | 4.22 | 0.058 |
| **Factor Interaction** | F | P | F | P |
| Gender * Co-contraction | 5.76 | 0.014* | 0.2 | 0.819 |
| Gender * Elbow Angle | 7.32 | 0.016* | 1.18 | 0.294 |
| Co-contraction * Elbow Angle | 1.64 | 0.227 | 4.22 | 0.035* |
| Gender * Co-contraction * Elbow Angle | 1.3 | 0.301 | 0.83 | 0.454 |

Women reached between 72-100% of male stiffness values and 42-76% of male damping values (see Figure 2.4 showing mean (SD) absolute values of stiffness and damping values). At the middle (51-70% MVC) co-contraction EMG level, men had 1.8~2.5 times higher absolute values of stiffness and 3.4 times higher absolute values of damping. Minimum and maximum values of non-normalized elbow stiffness were 25.8 Nm/rad and 191.7 Nm/rad in males, and 10.3 Nm/rad and 106.5 Nm/rad in females, respectively; the corresponding values for damping coefficients were 0.47 Nms/rad and 11.21 Nms/rad in males, and 0.38 Nms/rad, 3.33 Nms/rad in females, respectively.

The pattern of least squares means estimates (Figure 2.5) and the significant gender by triceps co-contraction (p = 0.014), and gender by initial elbow angle (p = 0.016) interactions (Table 2.2) suggest that stiffness was more affected by co-contraction level in males and by initial elbow angle in females. Gender differences were significant, especially at the highest (71-90% MVC) co-contraction level and in the most extended (10 degree) posture. However, no differences were found in damping coefficients between males and females.

In the six males and five females for whom strength data were available, the correlations between bilateral arm protraction strength and unilateral mean stiffness, K, and damping, B, values at the highest co-contraction condition (71-90% MVC) did not reach significance (Figure 2.6). While, a trend toward a gender difference in stiffness is observed (Figure 2.6, left), none is evident for damping (Figure 2.6, right).

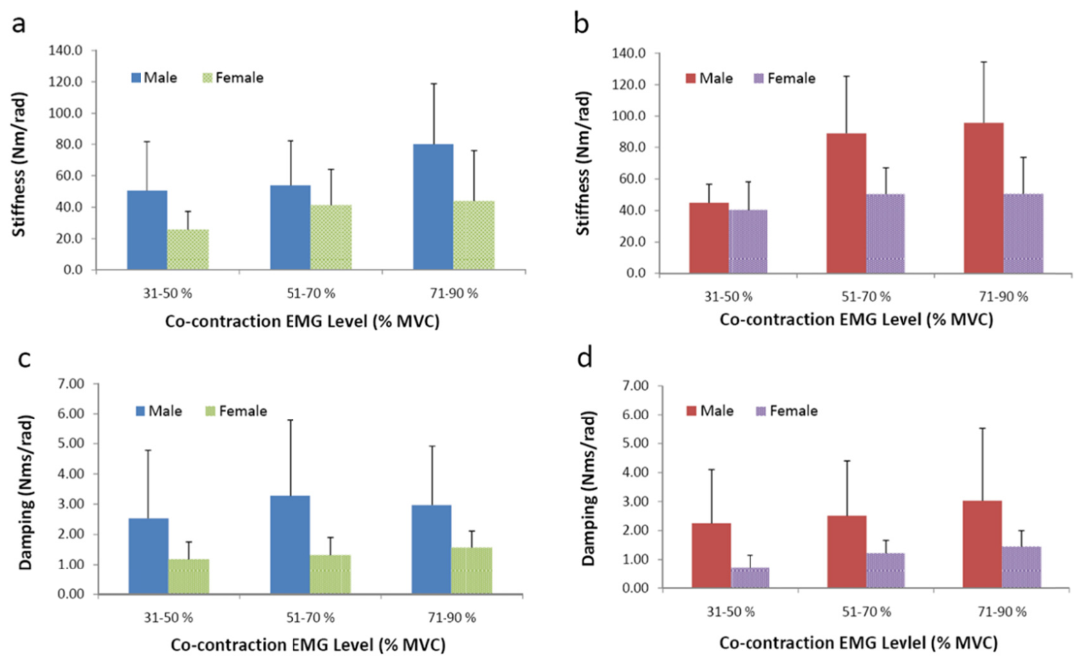


**Figure 2.4** Mean absolute values of elbow rotational stiffness (top panels: a, b) and damping coefficients (bottom panels: c, d) at an initial angle of 10 degrees (left panels: a, c) and 25 degrees (right panels: b, d) as a function of triceps co-contraction level. The bars denote one standard deviation.

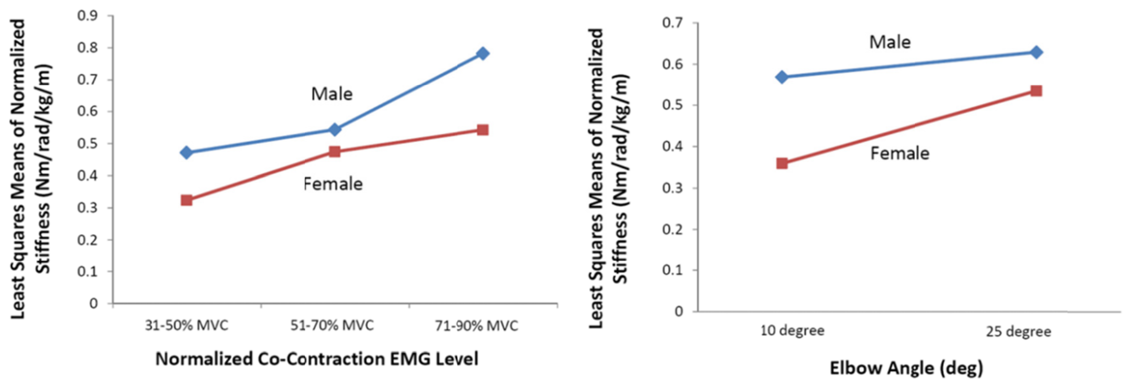


**Figure 2.5** Least squares means estimate for normalized stiffness suggesting gender/triceps muscle co-contraction interaction effect (left panel) and gender/elbow angle interaction effect (right panel).

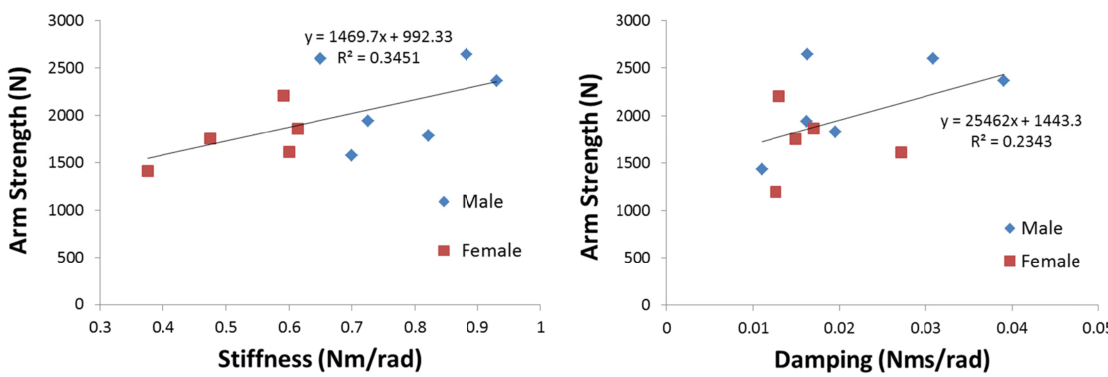


**Figure 2.6** Scattergrams showing the correlation between bilateral arm protraction strength and unilateral mean stiffness (left) and damping (right) parameter values.

## 2.5 Discussion

This study provides the first experimental evidence that the pre-activated elbow extensor musculature of males exhibits greater elastic and viscous resistance to stretch than females under a standardized step elbow flexion loading. This was true both in terms of absolute values as well as after normalization by body size (body weight times stature). This gender difference is consistent with the functional result of 12% less elbow deflection in males when they manually arrest an oncoming ballistic pendulum at chest level (DeGoede et al., 2002). These gender differences in upper extremity responses are consistent with similar differences in the resistance of the active musculature of the lower extremities to stretch, even in size-matched young men and women (Wojtys et al., 2003; Granata et al., 2002).

In the present study women exhibited 72-100% of male stiffness values and 42-76% of male damping values. These gender differences are consistent with the gender differences found in the literature for other joints. For example, the absolute active rotational knee stiffness in women has been found to be 56-73% of that in men, and female damping values ranged from 36-63% of male damping values (Granata et al., 2002). If the elbow extensor muscles, the triceps brachii, which are agonists during elbow extension, are assumed to be preactivated at the same volitional level, a reasonable explanation for the gender difference found in the present study is the greater arm muscle mass (see 2.2 Introduction) and elbow extensor strength of males (DeGoede and Ashton-Miller, 2003). The present findings in active muscle extend those of Lin et al. (2005) who studied the stretch behavior of passive human elbow muscle; however, the gender difference in normalized elbow muscle stiffness or damping in passive muscles did not achieve significance.

The present absolute values of elbow stiffness and damping are slightly higher than published values measured under a variety of testing conditions (Table 2.3) in upper extremities. This trend may reflect the higher muscle activation levels used in the present study. For example, it is the elbow stiffness values of 10.3 Nm/rad for the female subjects and 25.8 Nm/rad for the male subjects, measured at the lowest 'Normalized Co-contraction EMG Level' (Table 2.1), that correspond with published stiffness values. Our results also corroborate the published values that were obtained under 'comfortable' maximal co-contraction conditions (Popescu et al., 2003). On the other hand, our damping values were systematically larger than those reported by Popescu et al. (2003) particularly for fully activated male muscle (Table 2.3). This may again reflect differences in the levels of triceps muscle activation during the testing. Older women are less likely to successfully use their arms to arrest a fall, or they fail to appropriate use their arm to arrest the fall: both these behaviors carry a higher risk for head impact than males (Vellas et al., 1998). It is possible that with their age-related loss in strength, which they themselves instinctively may be aware of, older women fear flexion buckling of the elbow and thus are unable to avoid injury. They are caught on the horns of a dilemma. If they choose to avoid flexion-buckling by landing with a straight elbow joint then they risk a Colles fracture of the wrist; but if they permit flexion buckling of their elbow then they risk head or cervical spine injury (DeGoede et al., 2003). The present results show that this latter scenario is a reasonable fear. The lower maximal elbow extensor stiffness in women would lead one to predict a smaller capacity to prevent elbow flexion buckling under fall-related impact scenarios than in males of similar body size. This underlines the importance of maintaining elbow extensor strength with age. This is especially salient for those with obesity because the extra weight reduces the arm extensor strength-to-body weight ratio.

**Table 2.3** Ranges of non-normalized elbow stiffness and damping coefficients found in the literature.

| Stiffness ( Nm/rad ) | Damping ( Nms/rad ) | Movement Condition | Muscle Condition | Literature |
|---|---|---|---|---|
| 10.3 - 191.7 | 0.38 - 11.2 | voluntary motion | Active | Current Study |
| 17 - 55 | 0.5 – 1.8 | 0.5 Hz and 1 Hz voluntary motion | Active | Xu and Hollerbach, 1999 |
| 26.8 - 46.8 | - | reaching movement | Active | Flash, 1987 |
| 5 - 21 | - | ballistic, voluntary motion | Active | Gomi and Kawato, 1996 |
| 5 - 200 | 0.2 - 2.6 | voluntary extension motion | Active | Popescu, Hidler et al., 2003 |
| 42.9 - 168.8 | 3.92 - 9.09 | time-varying motion (catching a ball) | Active | Lacquaniti, Carrozzo et al., 1993 |
| 2.26 - 2.62 | 0.53 0.73 | pendulum test | Passive | Lin, Ju et al., 2005 |

This study has several limitations. One limitation of the study design is the relatively modest sample size which limited statistical power to less than 80% for the effect of gender on the damping coefficient (Table 2). A second limitation is that the study of gender differences would have been stronger had we recruited men and women of similar size (body height and weight) (Wojtys et al., 2003). Therefore, to counter bias due to the larger size of the males and their upper extremities (see Introduction), we only tested the three hypotheses using data that were normalized by body size. A third limitation is that the flexion moment due to the ~ 51 N step force was less than that would be induced by the 745-880 N end-load known to act on the hand/wrist during a natural fall arrest (DeGoede and Ashton-Miller, 2002). On the other hand, it was of a similar order of magnitude to that induced during a “stiff arm” fall arrest strategy (DeGoede and Ashton-Miller, 2002; DeGoede and Ashton-Miller, 2003), given that the elbow deflection was similar in both cases. This helps validate the present experimental design as being promising for testing middle-aged or even older adult elbow elastic and viscous properties.

A fourth limitation is the relatively weak correlation found between bilateral maximum arm protraction strength and maximum unilateral values of elbow extensor stiffness and damping (Fig. 6). Blanpied and Smidt (1993) found fairly strong linear correlations between ankle plantarflexion strength, stiffness and damping. One reason for our weak correlations could be the method for testing arm extensor strength: this relied on subjects having to stabilize their shoulder and wrist muscles in order to exert a maximum arm protraction effort. It is possible that the requirement for the subject to exert significant contractions of their shoulder adductor and flexor muscles, as well as their wrist flexor and hand power grip muscles, caused variability that weakened the correlations. In retrospect, less scatter might have been obtained had subjects exerted a unilateral single joint test of elbow extensor strength, rather than a bilateral multi-joint test of arm protraction strength. It is also possible that better correlations might have been found had we recruited only athletes because of their better body awareness, motor control and coordination abilities.

A fifth limitation was the use of just two different initial angles of elbow ($10^o$ and $25^o$) when a larger range of elbow initial angles has been observed in forward fall arrests (DeGoede and Ashton-Miller, 2002). Future studies might explore the effect of greater initial flexion angles to place the triceps muscle at longer fiber lengths. This would be expected to reduce the resistance found when the muscle is stretched beyond its short range stretch (Edman et al., 1978). However, one ethical concern is that larger stretches at higher contraction forces also increase the risk of injury (Brooks et al., 1995), so large stretches are contraindicated. While the subject only monitored triceps activity, and triceps stiffness can theoretically only affect the rising (elbow flexion) phase of the response in Figures 2 & 3, it is possible that the magnitude of biceps contraction might have affected the elbow stiffness on the descending (elbow extension) limb of that curve. However, there was no significant difference in the overall results whether one analyzed the ascending limb alone, or combined the ascending and descending limbs together. So it appears as if the triceps muscle activation and its mechanical properties dominated the measured elbow dynamic response. Lastly, joint stiffness can vary with time during a step response (Lacquaniti et al., 1993), so it might be worth investigating whether time-varying stiffness and/or damping constants would have given a better fit to the data than our assumption of constant stiffness and damping parameters.

## 2.6 Acknowledgments

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# CHAPTER 3

## On The Predicted Buckling Behavior of the Human Upper Extremity Under Impulsive End-Loading: Effects of Age, Gender, Muscle Stretch Behavior

### 3.1 Introduction

Falls are a leading cause of unintentional injury in all ages (CDC 2009-2018 and WISQARS$^{TM}$ (Web-based Injury Statistics Query and Reporting System)). When an arm is used to 'break' a fall, the impulsive end-load at the wrist can easily reach one body-weight (1*BW) or more (Burkhart and Andrews, 2013; Lo and Ashton-Miller, 2008; DeGoede and Ashton-Miller, 2002 & 2003; Dietz et al., 1981) which is enough to induce bone and joint fractures on a hard surface (Frykman, 1967). If the upper extremity gives way or buckles under such a load then there is a risk of the head striking the ground to cause traumatic brain injury (Chapter 1). In fact, fall-related head injuries are a problem in the elderly and they increase dramatically with age (Faul et al., 2010; Kannus et al., 2007; Jacobsson et al., 2007; Stevens et al., 2006; Hausdorff et al., 2001; Hornbrook et al., 1994).

The conditions under which a upper extremity buckles under an impulsive load are not well understood. We have previously reported an experimental study results on the effects of gender, the level of pre-cocontraction and initial elbow angle under a step increase of end load (Lee and Ashton-Miller, 2011). In that study we measured the elastic and viscous resistance of the actively contracting elbow extensor muscles in healthy young adults. Because of the risk of injury to older adults, an in silico approach might hold promise for studying the effect of age on arm buckling behavior.

Several computer simulation studies have been published on the biomechanics of falls. In the sagittal plane, Chiu and Robinovitch (1998) used a two-degree-of-freedom impact model to predict the impact loading on upper extremity in forward falls from different fall heights and suggested that fall heights greater than 0.6 m carry significant risk for wrist fracture. DeGoede and Ashton-Miller (2003) developed a five-link rigid-body model to investigate the effect of arm kinematics and upper extremity joint stiffness on the magnitude of hand impact force resulting from a partial forward fall. They predicted that older women with below-average bone strength risk a Colles fracture when arresting typical falls, particularly with an extended arm. Kim and Ashton-Miller (2009) also used a 2 degree-of-freedom discrete impact model to understand the dynamic responses in bimanual forward fall arrests. So in silico studies can be helpful for identifying risk factors for injury, but none of these models estimated the sensitivity of upper extremity buckling loads to biomechanical factors.

Most of these models focused on studying the dynamics at the instant of the impact without addressing how muscle contraction state affected arm kinematics and buckling load. The level of muscle pre-activation prior to the impact has been shown to closely correlate with the kinematic behavior of upper extremities when resisting a step increase in end-load (Lee and Ashton-Miller, 2011). However, the end-loading on an arm during a fall arrest is not a step increase in load but an impulsive loading. Therefore, the factors that determined the arm buckling behavior remain under an impulsive loading poorly understood.

When an arm is used to arrest a fall, the end-load typically forces the elbow into flexion. This causes the elbow extensor muscle to be forcibly stretched. Physiologists have studied how actively contracting muscle responses to a sudden stretch. As shown in Figure 3.1 [Left], a single muscle fiber on the plateau and decreasing regions of the sarcomere length-tension curve exhibited a high initial stiffness followed by a small stiffness to no force increase as the fiber continues to active stretch (Edman et al., 1982; Julian et al., 1979; Edman et al., 1978). This non-linear (bi-linear) stiffness response in active skeletal muscles has also been shown in whole muscle rabbit tibialis anterior muscle (in Figure 3.1 [Right], Grover et al., 2007), cat gastrocnemius muscle (Malamud et al., 1996), as well as human knee flexors (Shim and Garner, 2012) and extensors (Hahn et al., 2010). So it is possible that, for a given muscle contraction intensity, the shape of the force-length curve of an elbow and /or shoulder extensor muscle could affect the way the limb behaves under a sudden increase in end-load. Therefore we need to explore the effect of non-linear muscle stretch behavior on arm response to an impulsive end-load.

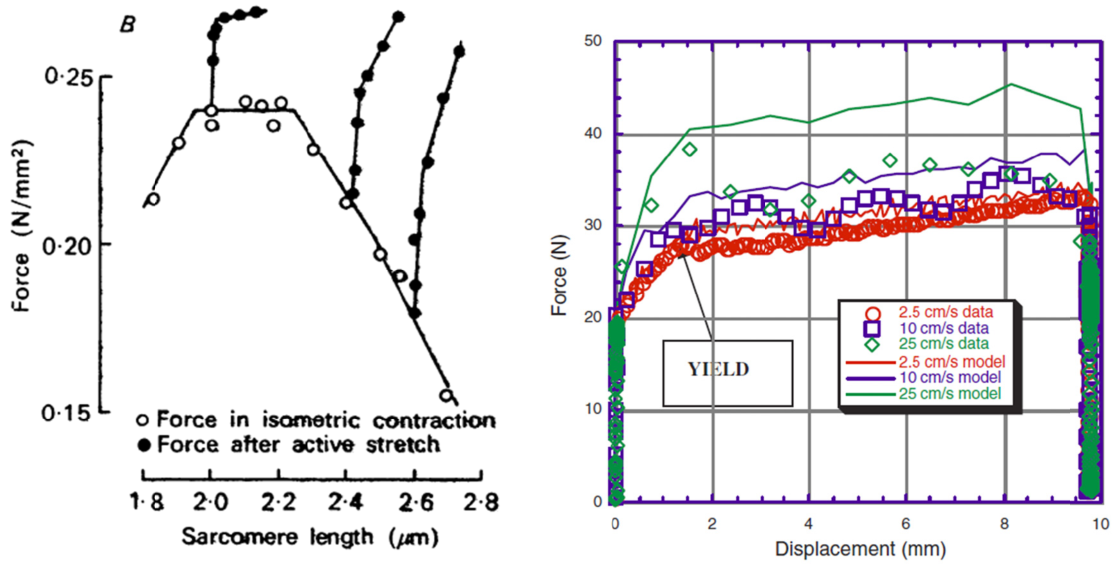



**Figure 3.1** [Left] Effect of increasing amplitude of stretch at three different starting points on the isometric length-tension curve in a single muscle (Edman et al., 1978). [Right] Typical force vs. displacement data for tetanized muscle and dynamic systems simulation (Grover et al., 2007).

There is a gender and age difference in the rate of fall and fall-related injuries. Older women experience significantly more falls and more injuries. For example, they sustain approximately four times as many distal radius and/or ulna fractures than do older men (Stevens, 2005). One of the explanations of age difference is the loss of strength with age, even in healthy and physically active older adults. For example, Lo (2006) used the joint torque data for gender difference and the age-related muscle strength decline for age difference to simulate arresting a forward fall. Anthropometry data including muscle strength and joint torque data by age and gender would be used to estimate the buckling behavior of upper limbs in this study.

Our working hypothesis is that there exists a magnitude of impulsive end-load that will buckle an upper limb for a given muscle precontraction state. Further, older age, female gender and lower precontraction levels in the arm protraction muscle will reduce that buckling load. So having sufficient arm extensor strength and pre-contracting the arm extensors above the threshold for arm buckling would appear to be important for avoiding head injury. In silico studies offer the possibility to explain these factors in a logical manner.

In this chapter we will use three computer simulation studies to address the above working hypothesis: (1) to evaluate the magnitude of impulsive end-load required to buckle the limb and determine how it depends on age, gender and the initial elbow angle (Lee and Ashton-Miller, 2008: Model I); (2) the effect of non-linear elbow extensor muscle force-length responses to a sudden stretch and elbow angle-dependent triceps moment arm on the arm deflexion behavior (Lee and Ashton-Miller, 2009: Model II); and (3) how the relationship between the magnitude of the impulsive ground reaction end-load end elbow and shoulder deflexion angles is affected by arm extensor contraction state and stiffness (Lee and Ashton-Miller, 2013: Model III).

## 3.2 Methods

### 3.2.1 Model I: Two dimensional (2-D) model of the sagittal behavior of a human arm to an impulsive end-load

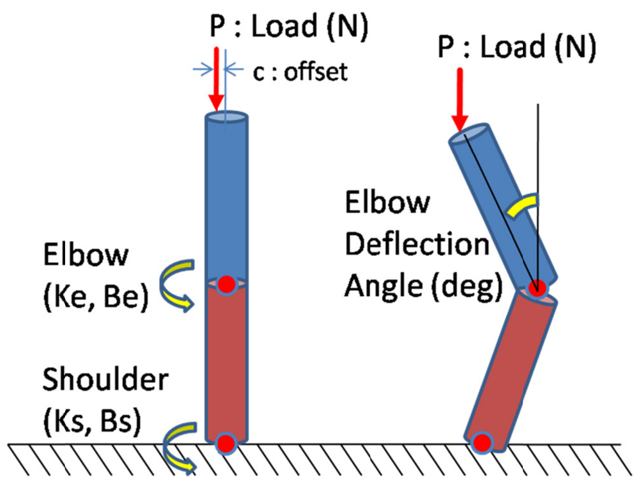



**Figure 3.2** (Model I) The inverted arm model in the sagittal plane. The bottom red dot represents the shoulder joint. The Model I was used to determine whether a buckling load exists to estimate the effects of elbow and shoulder stiffnesses ($K_E$ and $K_S$) and damping properties ($B_E$ and $B_S$).

A planar, two-link, lumped parameter, musculoskeletal model of the adult upper extremity was developed with frictionless revolute joints at the wrist, elbow and shoulder (Figure 3.2, Model I). Anthropometric properties and ground surface stiffness were taken from DeGoede and Ashton-Miller (2003). The resistance of the precontracted elbow extensor muscles to forced flexion was modeled with a rotational spring and damper at the elbow whose linear coefficients we identified from impulsive measurements in 5 females and 3 males (Mathias, 2006). These coefficients varied linearly with muscle pre-activation levels, as they do with muscle force (Blanpied and Smidt, 1993). The maximum volitional isometric shoulder flexor muscle strength, rotational stiffness and viscous coefficients were assumed to be 2.0-times and 1.5-times those for the elbow extensor muscles in healthy males and females, respectively (Stobbe, 1982). The model limb was flexed 10 degrees and end loaded by an axial impulsive force of known magnitude and time course (DeGoede and Ashton-Miller, 2002). Simulations were run in ADAMS 2005 R2 using 10 ms increments.

To simplify comparisons, we normalized the results to those involving the usual (70% of Maximum Voluntary Contraction, MVC) muscle pre-contraction levels measured in forward falls (DeGoede and Ashton-Miller, 2002) and we validated results using data from their experimental results (DeGoede and Ashton-Miller, 2002). We ran sensitivity analyses; compared to young males, size-matched young females (~25 years) and older males (~70 years) were assumed to exhibit a 0.2 decrement and older females (~70 years) a 0.4 decrement in their arm strength-to-weight ratios and stiffness and viscosity coefficients.

### 3.2.2 Model II: Two dimensional (2-D) model to examine how arm response is affected by the non-linearity of arm extensor muscle resistance to sudden stretch

A planar, two-link, lumped parameter, musculo-skeletal model of the adult upper extremity was developed using Adams 2008 R3 engineering software (Model II). Frictionless revolute joints were assumed at the wrist, elbow and shoulder. Segment anthropometric, mass and inertial properties and ground surface stiffness were taken from DeGoede and Ashton-Miller (2003). The resistance of the precontracted elbow extensor muscles to forced flexion was modeled with a rotational spring and damper at the elbow and shoulder identified from impulsive measurements in eight adults.

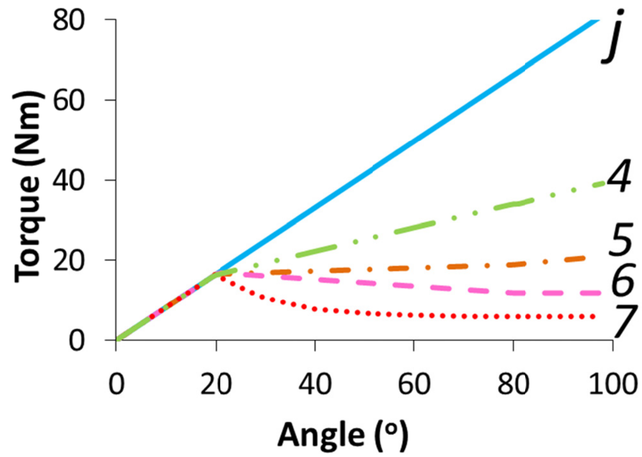



**Figure 3.3** Different elbow extensor muscle responses to a sudden stretch (curves 4 -7) for Model II: (a) both stiffness and damping coefficients decreased non-linearly with muscle force [curve 4], (b) a 'softening' relationship with a breakpoint occurring after the muscle had been lengthened by 14% (equivalent to 20° of elbow deflection) of its normal range of motion [curve 5], (c) a bilinear relationship, with the change in slope at 20° of elbow flexion [curve 6], and (d) a linear-exponential relationship [curve 7].

The short range muscle resistance to stretch was first assumed to vary linearly with muscle stretch force, but then the effects of non-linear relationships were systematically explored for larger stretches (Figure 3.3). The maximum volitional isometric shoulder flexor muscle strength, rotational stiffness and viscous coefficients were assumed to be 1.5-times those for the elbow extensor muscles in healthy males and females, respectively. Body weight (BW) and stature were assumed to be 713 N and 1.73 m in each gender group. The slightly flexed model limb (Model II) was end loaded by an axial impulsive load which was systematically varied between 400 N and 2.4 kN

(DeGoede and Ashton-Miller, 2003). At impact, the arm extensor muscles were considered to be isometrically precontracted at 70% of their maximum volitional activity (DeGoede and Ashton-Miller, 2002). Female gender was simulated by scaling male muscle stiffness and damping behavior by the gender factor $g$, where $g = 0.69$ to reflect the decrease in muscle stiffness and damping coefficients relative to size-matched young males (DeGoede and Ashton-Miller, 2003). Simulations were run for four initial pre-impact elbow flexion angles: $5^{\circ}$, $10^{\circ}$, $15^{\circ}$ and $20^{\circ}$ (where $0^{\circ}$ denotes elbow extension).

The sensitivity of arm load-deflection behavior to changes in extensor muscle responses to large stretches were examined as follows in Model II: (a) both stiffness and damping coefficients decreased non-linearly with muscle force [curve '4' in Figure 3.3], (b) a 'softening' relationship (Grover et al., 2007) with a breakpoint occurring after the muscle had been lengthened by 14% (equivalent to $20^{\circ}$ of elbow deflection, see Figure 1.9) of its normal range of motion [curve '5' in Figure 3.3], (c) a bilinear relationship, with the change in slope at $20^{\circ}$ of elbow flexion, [curve '6' in Figure 3.3] and (d) a linear-exponential relationship (Malamud et al., 1996), respectively [curve '7' in Figure 3.3]. These four 'softening' relationships were modulated by the 30% decrease in triceps moment arm from $20^{\circ}$ to $80^{\circ}$ flexion (Murray et al., 1995).

We also examined the effect of changing arm extensor muscle pre-activation level by multiplying the male muscle stiffness value of 1 by the muscle pre-activation ratio, $m$ in Model II. We also examined model sensitivity to changing the ratio of resistance provided by the elbow and the shoulder, respectively (see Table 3.1).

Model II outcomes included the increase in elbow angle, defined as the maximum change under load from the initial elbow angle. The extremity was considered to have buckled when elbow deflection angle reached 140°, and a head might strike the ground. OpenSim was used to estimate the triceps muscle sarcomere length and thence how much elbow flexion corresponded to the range of short range muscle stiffness.

**Table 3.1** Variable sensitivity simulation conditions. Muscle ratio (r) = gender factor (g) x muscle pre-activation ratio (m)

| Gender | Label in Figure 3.7 | | Initial Elbow Angle (°) | Ratio (r) = Gender Factor (g) x Muscle Preactivation Ratio (m) | |
|---|---|---|---|---|---|
| | | | | Elbow K & B | Shoulder K & B |
| Young Male for Figure 3.7 Upper | a | | 20° | 0.55 | 0.55 |
| | b | | 20° | 0.61 | 0.61 |
| | c | | 10° | 0.74 | 0.74 |
| | d | | 10° | 0.53 | 0.53 |
| | e | | 5° | 0.81 | 0.81 |
| | Deffrent Reduction of Elbow/Shoulder K & B Ratio | 1 | 5° | 0.81 (0%) | 0.32 (-60%) |
| | | 2 | 5° | 0.73 (-10%) | 0.41 (-50%) |
| | | 3 | 5° | 0.65 (-20%) | 0.49 (-40%) |
| | f | | 5° | 0.57 (-30%) | 0.57 (-30%) |
| Young Female for Figure 3.7 Bottom | h | | 20° | 0.55 | 0.55 |
| | i | | 15° | 0.55 | 0.55 |
| | j | | 15° | 0.69 | 0.69 |
| | Effect of Muscle Stretch Behavior (Figure 3.3) | 4 | 15° | Linear - Nonlinear Relationship | |
| | | 5 | 15° | 'Softening' Relationship with Breakpoint | |
| | | 6 | 15° | Bilinear Relationship (1/5) | |
| | | 7 | 15° | Linear - Exponentional Relationship | |
| | k | | 15° | 1.00 | 1.00 |
| | l | | 10° | 0.69 | 0.69 |
| | m | | 10° | 0.77 | 0.77 |

### 3.2.3 Model III: Three dimensional (3-D) model of arm and shoulder with lumped muscle representation to examine buckling behavior under an impulsive end-load

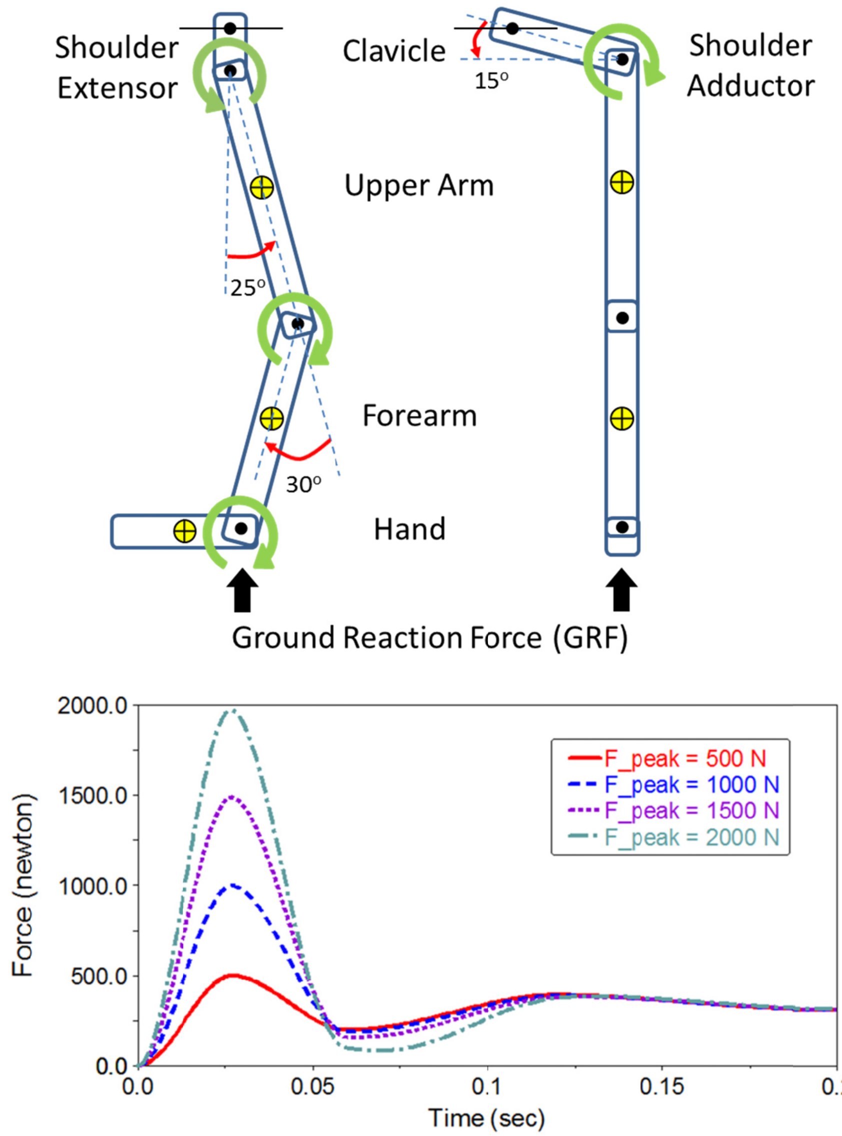



**Figure 3.4** [Upper] In silico Model III for simulating of the arrest of a sagittally-symmetric forward fall with the upper extremities (Left: sagittal plane, Right: axial plane). Black dots denote revolute joints at elbow and wrist joints, and spherical joints at shoulder clavicular-humeral joint, and at the sternoclavicular joint to ground. The green circular arrows represent lumped parameter elbow flexor and extensor muscles, and shoulder adduction and extensor muscles. They are represented by non-linear torsional springs with linear torsional dampers at each of the three joints. [Lower] Temporal histories of ground reaction forces for different peak loads ($F_1$) on the hand.

A 3-D, sagittally-symmetric, four-link (including hand, forearm, upper arm and clavicle), lumped parameter, musculoskeletal model of the adult upper extremity was developed using MD Adams$^{TM}$ 2010 engineering software (Model III in Figure 3.4). Segments were connected by two frictionless revolute joints representing wrist and elbow joints and by two frictionless spherical joints at shoulder and sternoclavicular joints. Segment anthropometric, mass, and inertial properties were taken from the literature (Lee and Ashton-Miller, 2011). Model height and weight were 1.78 m and 80 kgf.

The resistance of the precontracted elbow extensor muscles to forced flexion was modeled with a rotational spring and damper at the elbow whose linear coefficients we identified from impulsive measurements in 18 adults (9 females) (Lee and Ashton-Miller, 2011). We used these elbow data to identify muscular resistances (stiffness, K and viscosity, B) for the shoulder extensor (in sagittal plane) and adductor (in axial plane) muscles based on measured 3-D kinematics during these experiments (Figure 3.4).

Simulations of Model III were run with an initial pre-impact elbow flexion angle of $30^{\circ}$ (where $0^{\circ}$ denotes elbow extension) and shoulder extension angle of $25^{\circ}$ (where $0^{\circ}$ denotes the arm in the parasagittal plane (Figure 3.4). Model III simulations were run over the 200 msec post-impact at 1 msec increments. The sensitivity of arm load-deflexion behavior to changes in the peak load ($F_1$) of the impulsive ground reaction force on the hand with the usual muscle pre-contraction levels (70% MVC) were conducted from under 1*BW (785N) to 2.5 times BW in the peak force ($F_1$ = 500N – 2,000N) shown in Figure 3.4 bottom. We also conducted a design of experiments (DOE) analysis to examine the effect of variations in shoulder extensor and adductor muscle pre-activation levels ($K_{ext}$ = 1 – 8 Nm/deg for extensor stiffness and $K_{add}$ = 1 – 6 Nm/deg for adductor stiffness) and elbow muscles (elbow stiffness, $K_{elb}$ = 0.5 – 3 Nm/deg) on arm deflexion behavior. The effect of muscle viscosity, in units of Nms/deg, about a joint was assumed to have 1/10 the value of stiffness in Nm/deg (Lee and Ashton-Miller, 2011).

## 3.3 RESULTS

### 3.3.1 Model I: Finding Buckling load and Effect of Initial Elbow Angle

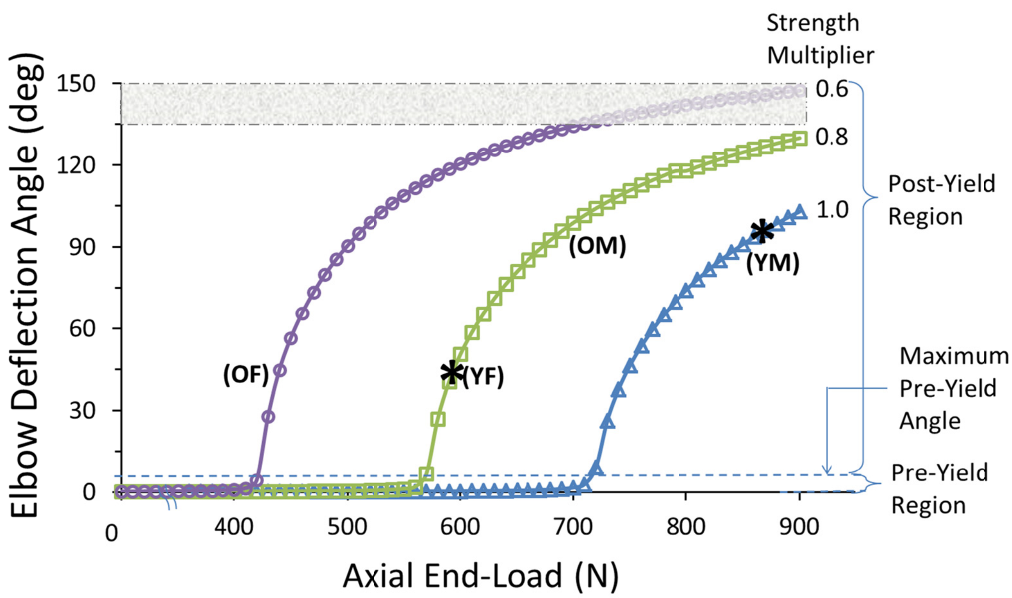



**Figure 3.5** Model I: Predicted elbow deflection behavior as a function of upper extremity axial end-load and 'strength multiplier' (age, gender or muscle pre-activity). The shaded region (at top) denotes limb collapse whereby the head would be predicted to strike the ground.

The buckling load of the healthy young male upper extremity was found to be 715 N (or just under 1*BW [73 Kgf]) (Figure 3.5) [with a greatest allowable elbow deflection permitted by young males (103 degrees, DeGoede, 2002) of 880 N ('YM' shown as a black star in Figure 3.5)] using Model I simulation. The corresponding buckling [allowable] loads for young female, older male and older female were 565 N [595 N], 565 N [700 N] and 420 N [440 N], respectively (Figure 3.5). The single black data point at 45 degrees on the YF curve showed the maximum average deflection that 10 healthy young female subjects actually permitted (45 degrees) in falls to the ground in our lab. This corresponds to a load of 600 N. The limb deflection behavior is divided into 'pre-yield' and 'post-yield' according to the elbow deflexion angle (Figure 3.5).

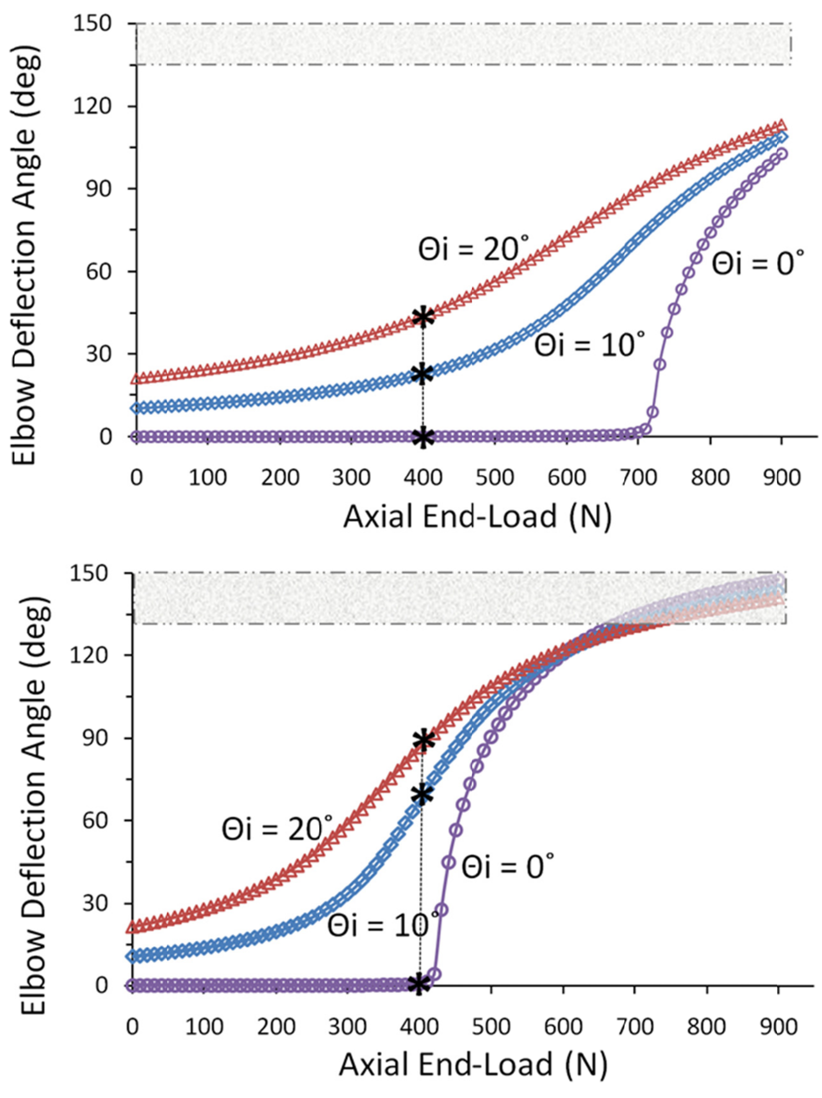
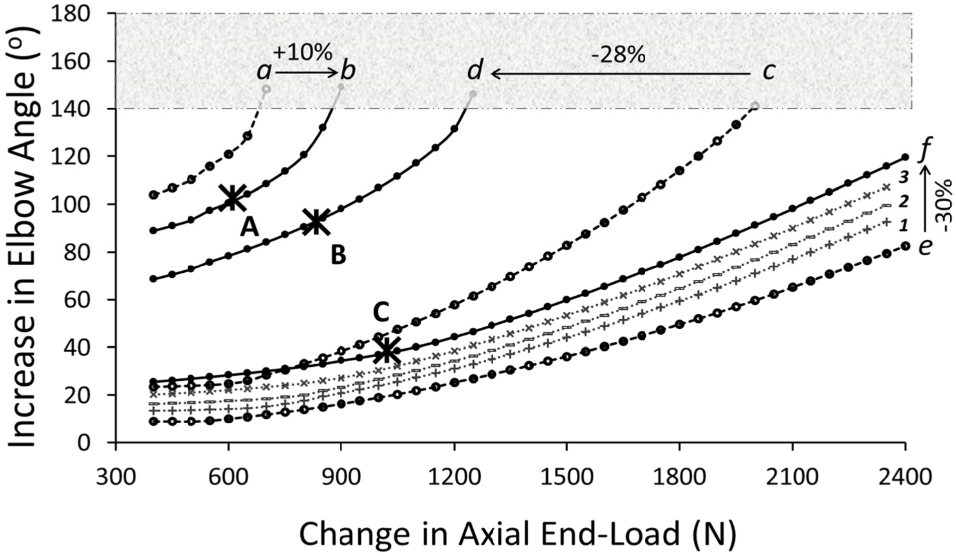



**Figure 3.6** Effect of initial elbow angle (0, 10 and 20 degree) on the limb deflection behavior of the young male limb (upper) and the older female limb (bottom) using Model I.

Initial elbow angle was found to significantly affect the buckling load of the extremity (Figure 3.6, Model I), but not the buckling load at which the head would strike the ground (i.e., 665 N in 'OF' in Figure 3.6 bottom). If we took a look at under 400 N end-load for example, as the initial elbow angle increase, the elbow deflection angle would be 20 degree and 40 degree for the young male (shown as three black stars in Figure 3.6 upper). However under the same load condition, the older female had much larger deflexion angles, 70 degree and 90 degree (shown as three black stars in Figure 3.6 bottom). This showed that the older female might strike their head easily than the young male.

### 3.3.2 Model II: Non-linear Muscle Stretch Responses

In the young male model II in Figure 3.7 upper graph, each curve was labeled with small letter *a-f* which corresponded to 6 different initial conditions (Table 3.1). Curve set *(a, c, e,* shown as dotted lines) are based on initial elbow angles of $20^\circ$, $10^\circ$, $5^\circ$ and values of the product r, where muscle ratio (*r*) = gender factor (*g*) x muscle pre-activation ratio (*m*), with values of 0.55, 0.75, 0.81, respectively in Table 3.1. Curve set *(b, d, f,* shown as solid lines) show how much *m* had to be modified by a multiplier to fit DeGoede's experimental data (DeGoede and Ashton-Miller 2002, three '⋇' points with capital letter A, B, and C). For example, for an initial $5^\circ$ elbow angle, *m* had to be modified from 0.81 to 0.57 (*e* → *f*), equivalent to a 30% decrease in both model elbow and shoulder stiffness. Likewise, a 28% decrease *(m = 0.74 → 0.53)* was needed to modify line *c* to *d,* and a 10% increase to modify line *a* to *b (m = 0.55 → 0.61)*.



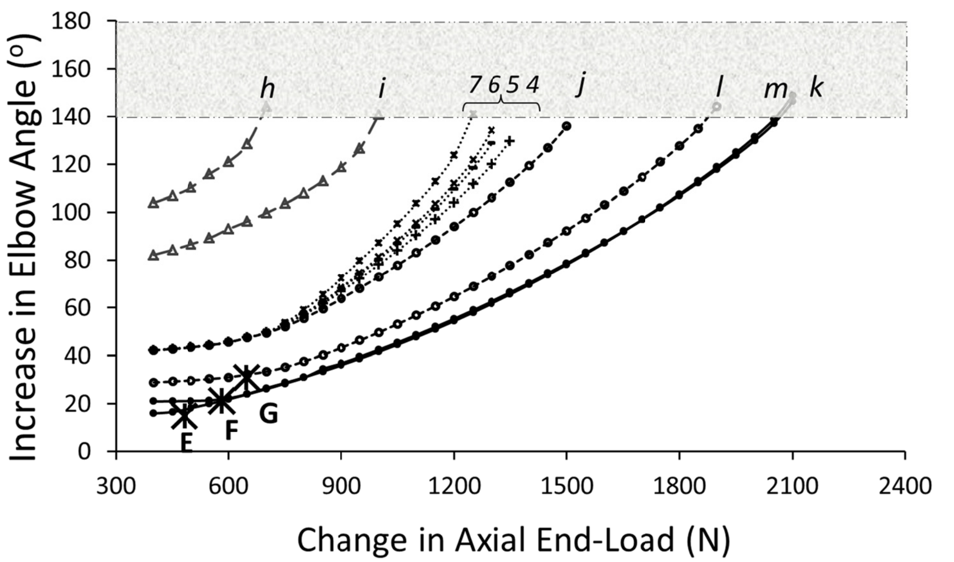




**Figure 3.7** Model II: Predicted elbow deflection behavior as a function of the impulsive increase in axial end-load, muscle tensile properties, and initial elbow angle in the Young Male (upper) and Young Female (bottom) limb. The shaded region denotes limb collapse. See text for abbreviations.

The effect of differentially reducing elbow rotational resistance relative to the shoulder resistance from that in curve *e*, can be seen in the gray dotted lines *(1, 2, 3),* where the elbow and shoulder reductions were 0% & 60%, 10% & 50%, 20% & 40%, respectively. Therefore, the buckling load is more sensitive to elbow than shoulder muscle stiffness. Buckling behavior was very sensitive to initial elbow angle at impact; increasing the angle, *f* $(5^\circ) \rightarrow d(10^\circ) \rightarrow a(20^\circ)$, caused the limb collapse load to decrease ~4-fold from ~2,400 N to 1,250 N to 700 N.

In the young female model II shown in Figure 3.7 lower graph, the three "Ж" data points are mean experimental data for young 5 females from Case et al., (2005) (Table 3.2). In the curve set (*j, k, l, m*), a 45% increase in *m* was needed to modify curve *j* to *k* so it passed through the correct data point; likewise a 12% increase (*l* → *m*). The curve set *(4, 5, 6, 7)* are the results of how softening relationships (see Methods 3.2.2 & Figure 3.3) modify the *j* curve which had an initial angle of $15^\circ$ and stiffness ratio of 0.69 to reduce the load for collapse in the young female. Curves *h* & *i* represent the loading cases in which the initial elbow angles were $15^\circ$ & $20^\circ$ with r = 0.55.

**Table 3.2** Comparison values between experimental conditions (DeGeode and Ashton-Miller, 2002 for the young male model and Case et al., 2005 for the young female model) and simulation conditions using Model II. The deflexion elbow angles ($\Delta\Theta$ in degree) as results were almost same.

| | Experimental Data | | | | | Simulation Data | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | Fall Height (cm) | Initial Elbow Angle Θi (°) | EMG | $F_1$ (N) | ΔΘ(°) | Initial Elbow Angle Θi (°) | Muscle Ratio ( *r* ) | $F_1$ (N) | ΔΘ(°) | Label in Figure 3.7 |
| Male | 100 | 174° | 0.81 | 1021 | 39 | 175° (5°) | 0.59 | 1021 | 38 | A |
| | 100 | 168° | 0.74 | 832 | 93 | 170° (10°) | 0.53 | 832 | 93 | B |
| | 100 | 163° | 0.55 | 611 | 103 | 160° (20°) | 0.61 | 611 | 102 | C |
| Female | 90 | 168° | 0.7 | 648 | 31 | 170° (10°) | 0.69 | 648 | 31 | G |
| | 80 | 168° | 0.7 | 584 | 21 | 170° (10°) | 0.77 | 584 | 22 | F |
| | 70 | 166° | 0.7 | 482 | 15 | 165° (15°) | 1.00 | 482 | 17 | E |

### 3.3.3 Model III: Finding Buckling Load in 3-D Model Including Elbow and Shoulder

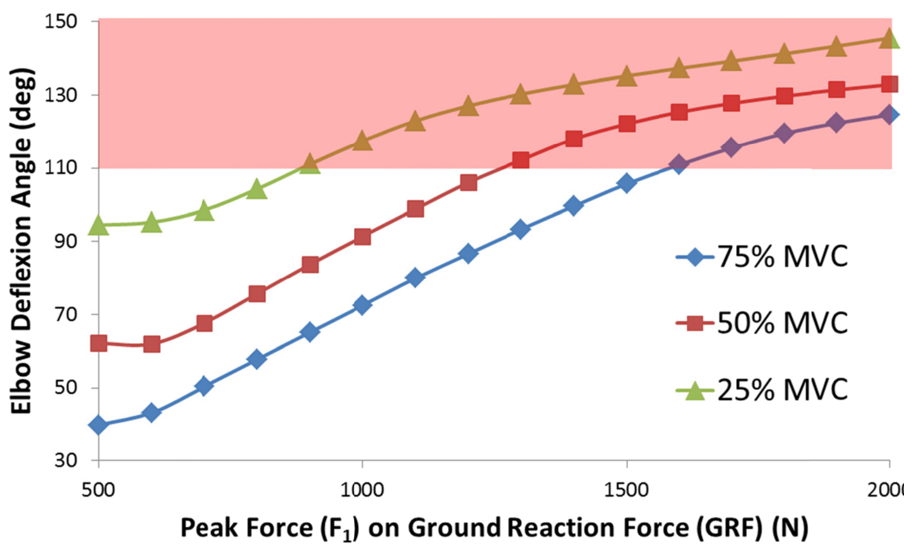



**Figure 3.8** Model III: Predicted buckling behavior as a function of upper extremity distal peak load ($F_1$) and elbow and shoulder muscle pre-contraction level (% MVC). The shaded region ($\Delta\Theta_{elbow} > 110°$) denotes limb buckling.

The limb was predicted to buckle when the peak ground force exceeded 1,600 N for a 75% MVC elbow and shoulder muscle pre-contraction level, 1,300 N for 50% MVC, and 900 N for 25% MVC, respectively (Figure 3.8). The peak force ($F_1$) is known to be related to fall height and muscle pre-contraction levels (DeGoede and Ashton-Miller, 2002). We found earlier that the stiffness and viscosity acting about a planar elbow joint markedly affected its buckling behavior (Lee and Ashton-Miller, 2011) and the same trend was found in this study. However, in this study, the shoulder adductor muscle properties significantly affected the shoulder adduction angle as well as the elbow deflexion angle, but neither the shoulder extensor muscle properties nor the shoulder adduction angle affected the final elbow angle (Table 3.3).

**Table 3.3** Model III sensitivity results shown as angle changes in percentage by varying muscle properties. The percentage to the left of the slash mark indicates the effect of a minimum modification in the parameter and the percentage to the right a maximum modification (see Methods).

| Variable | Elbow Deflexion | Shoulder Extension | Shoulder Adduction |
|---|---|---|---|
| **$F_1$** | -72% / +82% | -48% / +21% | -6% / +11% |
| **Elbow K & B** | -80% / +43% | -10% / +40% | -15% / +2% |
| **Shoulder Extensor K & B** | -0% / +0% | -53% / +50% | -0%/ +0% |
| **Shoulder Adductor K & B** | -13% / +5% | -43% / +10% | -50% / +14% |

## 3.4 Discussion

We simulated the behavior of upper extremity under impulsive end-loading in silico using three increasingly complex computer models: from a simple 2-D linear muscle model (Model I), a 2-D non-linear muscle model (Model II), and a 3-D non-linear muscle model (Model III) in a fall arrest scenario. In each of these simulation studies we estimated the buckling (critical) load, the effect of age, gender and pre-contraction of elbow and shoulder muscles, and the effect of muscle stretch properties on the responses of upper extremities in a forward fall arrest.

The first 2-D model simulation study (Model I) was a first estimate of the buckling load of a slightly flexed extremity in arresting a fall. In this Model I simulation study, the post-yield region behavior was conservative (may be overestimated) in that post-impact muscle inhibition was not considered. The ‘strength multiplier’ results in Model I (Figure 3.5) showed that buckling loads decrease roughly linearly with a reduction in muscle pre-activity, and vice versa. Despite a number of simplifying assumptions (as discussed in the last part of Discussion Section), we believed this simple model makes reasonable predictions because the behavior of the young adult model extremity matches that we had measured in actual forward falls (DeGoede and Ashton-Miller, 2002).

These results of the effect of older age underline the importance of older individuals maintaining as much arm protraction strength as possible. Given that the clinical studies (Sran et al., 2010; Chiu and Robinovitch, 1998) that the strength of the triceps muscle was a major contributor to upper extremity energy absorption and further that older women were substantially less able than young women to absorb energy in their upper extremities during arresting a fall simulation, age adversely affect upper extremity buckling loads. This could help explain why fall-related head injuries are more common in older females (for example, Watson and Mitchell, 2011; Fuller, 2000).

In the second simulation study using Model II, we concluded that upper extremity buckling was likely to occur in (a) males when landing with excessive initial elbow flexion (>20°) or inadequate elbow muscle pre-activation (<53% MVC); and in (b) females when landing with excessive initial elbow flexion (>20°), too little shoulder extensor muscle pre-activation (<30%), or elbow extensor muscle weakness (<58% MVC).

Also, we found that the muscle stretch behavior affected the behavior of upper extremities in elbow deflexion angles which was closely related with the limb buckling using Model II simulation study. The large non-linearity of how muscle softening relationships reduced the arm buckling loads. This model II study suggested women preactivate and stiffen their shoulder muscles more than males, and the buckling load was more sensitive to elbow than shoulder muscle stiffness.

We found that the greater initial elbow angle the greater the elbow deflexion angle. But when we considered that the effect of arm predictions may be an underestimate of this effect because the model did not take into account the length-term relationship of the elbow extensors which used. This would be expected to reduce the resistance found when the muscle was stretched beyond its short range stretch (Edman et al., 1978).

The third musculoskeletal Model III showed that limb buckling was sensitive to arm and shoulder muscle pre-contraction intensity, as well as the peak impulsive load acting on the limb. Because the shoulder adductor muscle state proved as important as the elbow extensor muscle state, it would wise to maintain good shoulder adductor strength, as well as triceps strength, if one is to prevent the head from striking the ground in a fall.

The present simulation studies include several limitations. The first and second 2-D models (Model I and Model II) used to simulate forward falls, which was the most common fall direction, and using upper extremities to arrest a forward fall is a common strategy (Hsiao and Robinovitch, 1998), could not capture out-of-sagittal-plane motions and joint torques. Many forward falls are not completely sagittally symmetric therefore this simplification would inevitably introduce some degree of inaccuracy to the model prediction.

In the first simulation, the Model I had several simplifying assumptions including the fact that the stiffness, K and damping, B for representing rotational muscles at elbow and shoulder were considered constant through the range of motion. So, the effect of varying muscle synergies was not considered, nor was the whole body dynamic effect.

In order to overcome above limitations, we explored the effect of non-linear triceps muscle stretch responses using Model II. However the second version model (Model II) was limited to a planar representation and did not consider posterior translation of the torso. This was considered in the third model (Model III) but the biomechanical representation of the shoulder was highly simplified compared to in a real fall scenario in which many shoulder girdle muscles than shoulder adductor and extensor muscles. In spite of the several assumptions and limitations, we believed the model studies captured the essence of arm behavior under impulsive end-loading. The findings that age, gender, and muscle pre-activation levels including linearity and non-linearity properties helps explain why fall-related head injuries increase with age. To prevent fall-related injuries, we would maintain good elbow and shoulder strength.

### 3.5 Acknowledgments

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# CHAPTER 4

## *In Vivo* Determination of Arm Muscle Stiffness and Damping Properties In an Impulsively End-Loaded Upper Extremity: Age, Gender and Pre-Contraction Level Effects

### 4.1 Introduction

The upper extremities are typically the first line of defense to protect the head and torso while arresting a fall to the ground (DeGoede et al. 2003). Impulsive loads on hand can reach 2–3 times body weight (BW) for a fall from even half standing height (DeGoede and Ashton-Miller 2002; Lo et al. 2008) and 1–4 kN for a fall from standing height (Dietz et al. 1981), enough to cause wrist fracture on a hard surface (Frykman 1967). To prevent a slightly flexed limb from buckling under an impulsive load applied to the hand (DeGoede et al. 2002; DeGoede and Ashton-Miller 2002; Lee and Ashton-Miller 2011), the arm muscles are usually pre-contracted to brace the arm for impact (DeGoede and Ashton-Miller 2003; Lo et al., 2003; Troy and Grabiner, 2007; Lo and Ashton-Miller 2008). Muscle pre-cocontraction may be required (Brown and Loeb, 2000) because the impulsive ground reaction force on the hand peaks rises so quickly (Dietz, 1981; Hsiao and Robinovitch, 1998; DeGoede and Ashton-Miller, 2002; DeGoede et al., 2003).

Because falls are a leading cause of TBI (traumatic brain injuries), particularly among the elderly over the age of 70 years (Ingebrigtsen et al., 1998; Kannus et al., 2001), clearly the protective use of the upper extremities was ineffective in these cases. This type of protective hand use is associated with a potential risk for wrist injury, a risk-benefit ratio that seems reasonable given the potential severity of head or hip injury in the absence of such a strategy. Vellas et al. (1998) demonstrated that the part of the body receiving the main impact in a fall was, in order of frequency: the hand (50% males, 33% females) and buttock (18% males, 24% females), followed by the head, knee, and arm. In older women, the most common fall-related fracture sites are the upper extremity, the hip, and the trunk or neck in that order (Sattin et al., 1990). For older men a similar pattern is observed although the fracture rates are halved. Therefore, a better understanding of the effects of how to properly use upper extremities to arrest a fall is necessary.

Firstly, we need to understand the behavior of the upper extremities under impulsive end-loading, especially the effects of advancing on age on the behavior of the arms under such large and violent loads. Case et al., (2005) had examined gender differences in upper extremity kinematics and impact loading in forward falls for young female adults to compare with the previous studies for young male adults (DeGoede and Ashton-Miller, 2002; Lo et al., 2003). They found that both genders use similar initial elbow and shoulder angle at impact, but the young women's post-impact deflexion angles were 4-times less at the elbow and 2-times less at shoulder extension than in the young men. This suggests that the young women were not willing to risk their arms inadvertently collapsing under the end-load. Lee and Ashton-Miller (2011) (Chapter 2) studied the factors which determined the arm buckling load and found that young female adults had lower rotational stiffness and damping resistance at elbow than young male adults of the same body size. That study was limited to a single joint, the elbow, and the end-loads applied to the hand not physiologic, being modest step increase in load.

Computer simulations suggest that shoulder muscle rotational stiffness and damping properties affect the behavior of upper extremity under impulsive load (Chapter 3). The anterior deltoid and pectoralis major play a role in preventing under shoulder when arresting a forward fall, while the elbow triceps brachii are the main muscles that determine the stiffness and damping properties of the elbow (Lee and Ashton-Miller, 2011). Without stereo radiography it is difficult to measure the shoulder deflexion angles during impact because the complexities of 3-D clavicular and scapular movements are hidden below layers of muscles. We therefore developed a lever arm drop-weight apparatus to study the movement of the shoulder in two planes: shoulder flexion/extension in the sagittal plane and shoulder ad-/abduction in the frontal plane.

In the Introduction (Chapter 1) and in the Chapter 3 simulations we studied how age, gender and pre-contraction level affect the dynamic response of an arm model to an impulsive end-load. The goal of this paper, therefore, was to test the primary hypotheses in healthy adults that neither gender, age, nor level of pre-cocontraction affect shoulder and elbow muscle viscoelastic properties (rotational stiffness and rotational damping coefficient) of the upper extremity under an impulsive end-load. The secondary hypothesis was elbow and shoulder stiffness and damping is proportional to muscle strength. We used computer simulation to calculate the stiffness and damping coefficients from the kinematic and kinetic data using an inverse dynamics optimization algorithm.

## 4.2 Methods

A full description of the methods has been published in Lee and Ashton-Miller (2013). Thirty eight healthy men and women [ten young males of mean (SD) age: 25.5 (2.7) years; 8 young females: 24.5 (3.1) years; 9 old males: 69.4 (3.4) years and 11 old females: 67.7 (2.4) years] participated in the study with written informed consent. Mean height and mass for the young males were 1.795 (0.077) m and 75.88 (6.74) kg, for the young females were 1.683 (0.060) m and 60.96 (7.67) kg, respectively and for the old males were 1.734 (0.085) m and 74.73 (11.83) kg and for the old females were 1.623 (0.039) m and 59.20 (7.01) kg, respectively.

Each subject was asked to lightly exercise his/her arm and shoulder muscles by doing several push/pull ups and various stretches against wall. We placed surface electromyographic (EMG) electrodes (Trigno™ Wireless System, Delsys, Inc., Boston, MA. USA) on the skin over the mid belly of selected arm muscles to measure non-dominant arm muscle activity. We then measured subject's resting and maximum voluntary pre-cocontraction (MVC) EMG levels of the triceps brachii (long and lateral head), biceps brachii (short head), the anterior deltoid, pectoralis major and serratus anterior muscles during elbow and shoulder flexion, extension and ab- and adduction by pulling up or pushing down on an handle attached to a vertical cable in series with an uniaxial force transducer (TLL-500, Transducer Techniques, Temecula, CA. USA) for measuring muscle strength.

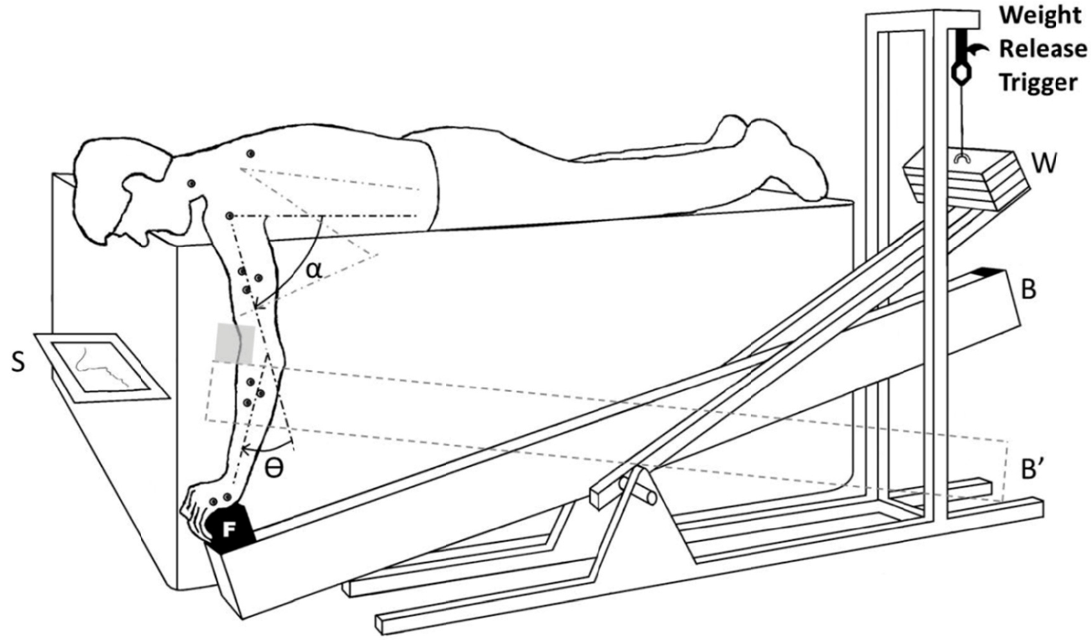



**Figure 4.1** Schematic of testing apparatus for the impulsive end loading test of a human upper extremity. Each subject lay on a padded table with left hand positioned on a force transducer (F). The subject was asked to concentrate on monitoring EMG biofeedback from his/her elbow extensor muscle activity on a display screen (S) and maintaining it at a certain level of effort. A weight (W) of 23 kgf was released onto the end of the lever-arm (B) by a remote trigger, applying an impulsive force to the wrist (at the other end of lever arm) causing elbow flexion and shoulder adduction (the end of the lever-arm position being changed from B to B'). Alpha (α) and theta (Θ) represent shoulder extension and elbow flexion angles, respectively.

Next, each subject was asked to lie prone on a table with the left wrist positioned on a 6 axis force transducer (MC3A-1000, AMTI, Newton, MA. USA) mounted at one end of a 76 mm x 152 mm x 2,032 mm hollow aluminum beam having a rectangular cross-sectional shape and wall thickness 6.35 mm (Figure 4.1). The beam was pivoted at its midpoint about a fulcrum formed from a pair of collinear needle bearings and mounted on an axle in the horizontal plane. An Optotrak Certus camera (Northern Digital, Inc., Waterloo, Canada) was to measure the displacements of arm, shoulder and neck optoelectronic markers (shown as dots in Figure 4.1) taped to the skin over wrist, elbow and shoulder joint bony landmarks.

When ready, the subject was asked to hold his/her hand "lightly" in contact with the force transducer and to concentrate on monitoring EMG biofeedback from his/her lateral head of the triceps muscle activity on a display screen and maintaining triceps activity either at rest, or 25, 50, and 75% MVC values from the main agonist muscle. A weight of 23 kgf was released (shown as "W" in Figure 1) from a height of 720 mm to impact the top surface of the other end of the beam, thence applying an upward impulsive force to the wrist via the force transducer, thereby causing elbow flexion, shoulder extension/adduction and trunk extension. The subject was instructed "not to intervene" before, during and after the weight drop. For example, if the trial was conducted with 50% MVC, the subject was instructed to contract the target muscle steadily at 50% of MVC during the test. Three trials at least were conducted at each of the three levels of muscle activation, and these were presented in randomized order.

3-D kinematics data were measured at 280 Hz from 15 infrared-emitting diodes adhered to the skin with double-sided tape. The kinematics and force data were digitally low-pass filtered (MATLAB, The MathWorks, 4th order Butterworth) with cutoff frequencies of 30 and 300 Hz. Surface electromyography (EMG) data were collected at 4 kHz. A band-pass 6th order filter with breakpoints at 40 Hz and 500 Hz was used to attenuate any movement artefacts in the EMG signal. Then, the data were digitally low-pass 4th order Butterworth filtered with a cutoff frequency of 30 Hz. EMG data were normalized by maximum MVC values. The muscle preactivation state for each trial was determined as the mean value of a 100 msec time window 50 msec before the weight drop.

### 4.2.1 Statistical Analyses

Descriptive statistics were undertaken for calculating joint marker kinematics, forces, and torques and EMG levels. A repeated measures analysis of variance (rm-ANOVA) was used to test the null hypothesis for age, gender, and three different muscle pre-cocontraction levels using SAS 9.3 software. A p-value of less than 0.05 was considered statistically significant for the three main effects (primary hypothesis). A Bonferoni correction was used for the interactions.

### 4.2.2 Inverse Dynamics Optimization Model

An inverse dynamics optimization algorithm was used to estimate the rotational stiffness and damping at the elbow and shoulder joints using the experimental applied flexion moment (torque), the limb inertia, joint angle, vertical displacement using MD Adams$^{TM}$ (MSC. Software Corporation, version 2010). A 3-D, sagittally-symmetric, four-link (including hand, forearm, upper arm and clavicle), lumped parameter, musculoskeletal representation of the proportion to height and weight of each subject was used as described in the next paragraph.

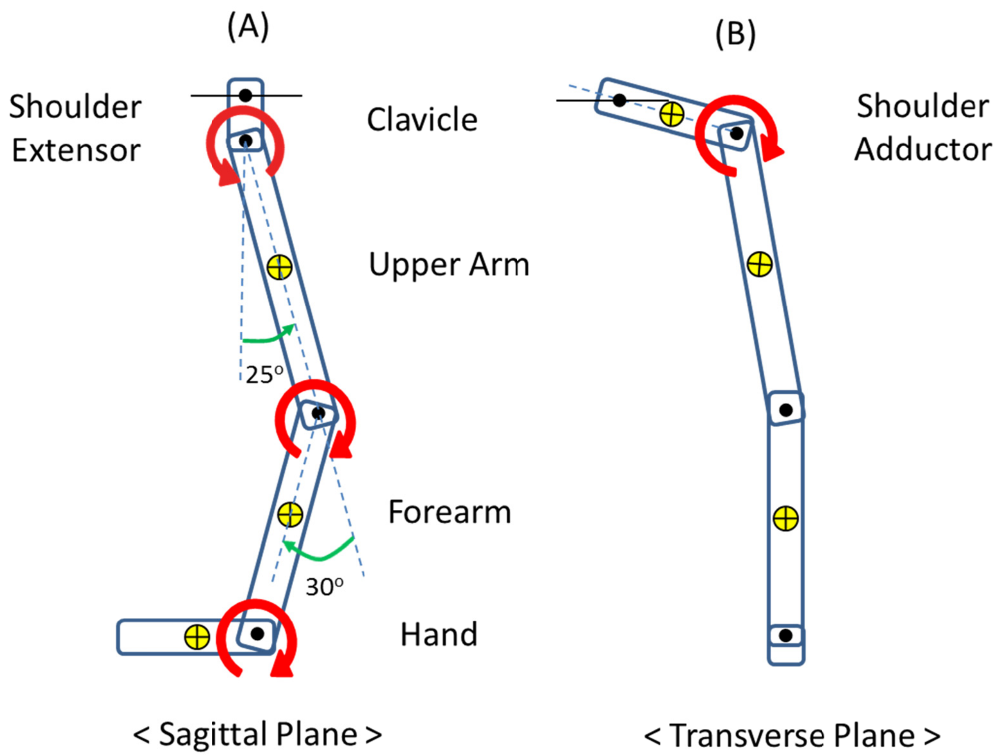



**Figure 4.2** *In silico* model for optimizing of the tensile stiffness (K) and damping coefficients (B) on the upper extremities (Left: sagittal plane, Right: transverse plane). Black dots denote spherical joints at wrist, elbow, shoulder and sternoclavicular joint for extensor and adductor, and at the clavicle to ground.

Arm segments were assumed to be connected by four frictionless spherical joints at wrist, elbow, shoulder and sternoclavicular joints in Figure 4.2. Segment anthropometric, mass, and inertial properties were proportionally assigned based upon the literature (Winter, 2005). The model arm muscles were represented by a bilinear torsional spring and linear damper in parallel at the elbow, and again at the shoulder and sternoclavicular joints. The bilinear behavior for each joint was characterized using two tensile stiffnesses ($K_1$, and $K_2$) and a single damping coefficient (B) in an optimization algorithm. Joint kinematics, including wrist angle, elbow angle, shoulder angle and wrist displacement, were measured as the arm was end loaded in a proximal direction with a time history measured in all trials of impulsive loading tests. Next, an optimization routine that minimized the square of the paired differences between corresponding points on the desired and calculated (measured) joint torques was processed to find $K_1$, $K_2$ and B of each joint. The values of Ks and B were normalized by subject body weight times height to reduce the effect of body size as a confounder in inter-subject comparisons and make the data more transferrable to other subjects.

$$Min\left\{Error = \sqrt{\sum_{Joint}(Torque_{Desired} - Torque_{Measured})^2}\right\}$$

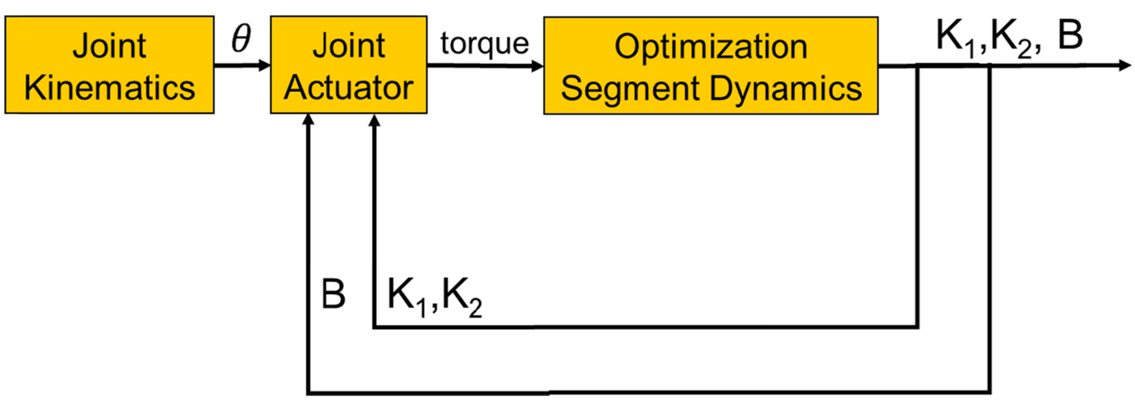



**Figure 4.3** Block diagram for showing the optimization algorithm among the components of muscle-joint model.

## 4.3 Results

### 4.3.1 Elbow and shoulder rotational stiffness and damping coefficient values

Sample flexion moment (torque) elbow and shoulder response for an impulse load in a subject YMWK is shown as the broken lines in Figure 4.4. The computer simulation optimized the variables of $K_1$, $K_2$ and B on each joint then the response of model with these values was plotted as solid lines in Figure 4.4. Table 4.1 shows the normalized mean values of elbow and shoulder rotational stiffness and damping found during 150 msec after an impact for each gender, age and pre-cocontraction levels. Across all subjects, the mean (± SD) of the first ($K_1$) and the second ($K_2$) stiffness were 0.843 (± 0.605) Nm/rad/kg/m and 0.860 (± 0.321) Nm/rad/kg/m for elbow, 1.105 (± 0.663) Nm/rad/kg/m and 0.135 (± 0.126) Nm/rad/kg/m for shoulder on transverse plane, and 9.904 (± 2.795) Nm/rad/kg/m and 0.788 (± 0.422) Nm/rad/kg/m for shoulder on sagittal plane, respectively. For the damping, values of 0.037 (± 0.032) Nms/rad/kg/m were found for the elbow, 0.012 (± 0.013) Nms/rad/kg/m for the shoulder in the transverse plane, and 0.097 (± 0.080) Nms/rad/kg/m for the shoulder in sagittal plane, respectively.

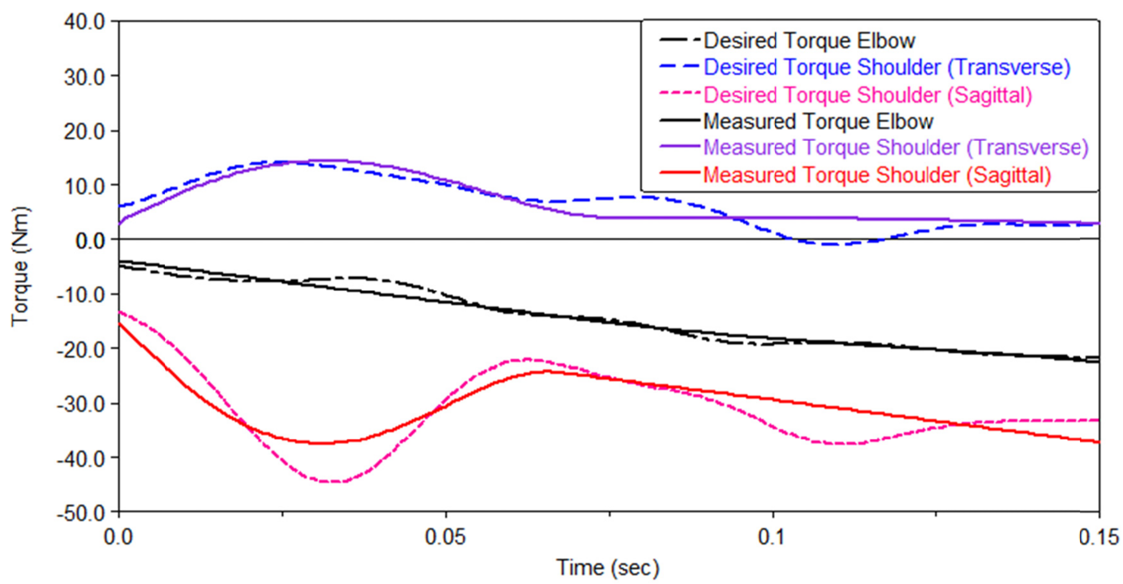



**Figure 4.4** Time course for showing the optimization result between the desired torque (broken lines from experimental result) and the measured torque (solid lines from computer simulation result) among the elbow and two shoulder joints from subject YMWK with pre-cocontraction level of 50% MVC.

Older adults had 57% − 95% and 76% of young adult stiffness and damping values; women had 51% – 80% and 66% of men's stiffness and damping values at elbow, respectively (see Table 4.1 showing mean (SD) absolute values of the first stiffness ($K_1$) and damping values (B)). We did not find significant differences in old vs. young, female vs. male in the normalized stiffness and damping values of the shoulder in either the sagittal or the transverse plane. However we found that the first stiffness coefficient ($K_1$) of the shoulder in the sagittal plane in female adults was significantly higher than in male adults; this finding will be discussed further in next section.

In testing of the primary hypothesis, Table 4.2 shows the main effect of age, gender, and pre-cocontraction level and the interaction for each joint stiffness and damping values ($p < 0.05$). The ANOVA (Table 4.2) demonstrated that pre-cocontraction level significantly affected normalized joint stiffness and damping coefficients. Age affected only the first stiffness coefficient for the elbow ($K_1$) and gender affected both the first stiffness ($K_1$) at the shoulder in sagittal plane and the damping coefficient (B) at the shoulder in the transverse plane. Stiffness and damping coefficients for the elbow and shoulder were proportional to higher levels of pre-cocontraction. However, women had a larger mean normalized stiffness value at the shoulder in both planes (Table 4.1).

**Table 4.1** Mean (SD) normalized stiffness and normalized damping coefficients of all subjects by age, gender, and level of pre-cocontraction.

| Elbow | *Normalized Stiffness (Nm/rad/kg/m)* $K_1$ | | *Normalized Stiffness (Nm/rad/kg/m)* $K_2$ | | *Normalized Damping (Nms/rad/kg/m)* $B$ | |
|---|---|---|---|---|---|---|
| | Mean | (SD) | Mean | (SD) | Mean | (SD) |
| Age | | | | | | |
| Old | 0.625 | (0.538) | 0.840 | (0.306) | 0.032 | (0.031) |
| Young | 1.099 | (0.585) | 0.882 | (0.341) | 0.042 | (0.033) |
| Gender | | | | | | |
| Female | 0.562 | (0.496) | 0.759 | (0.275) | 0.029 | (0.030) |
| Male | 1.109 | (0.582) | 0.955 | (0.335) | 0.044 | (0.032) |
| Pre-cocontraction | | | | | | |
| 25% MVC | 0.709 | (0.552) | 0.824 | (0.328) | 0.027 | (0.027) |
| 50% MVC | 0.855 | (0.644) | 0.855 | (0.326) | 0.038 | (0.033) |
| 75% MVC | 0.965 | (0.606) | 0.899 | (0.315) | 0.045 | (0.035) |
| All | 0.843 | (0.605) | 0.860 | (0.321) | 0.037 | (0.032) |

| Shoulder in transverse plane | *Normalized Stiffness (Nm/rad/kg/m)* $K_1$ | | *Normalized Stiffness (Nm/rad/kg/m)* $K_2$ | | *Normalized Damping (Nms/rad/kg/m)* $B$ | |
|---|---|---|---|---|---|---|
| | Mean | (SD) | Mean | (SD) | Mean | (SD) |
| Age | | | | | | |
| Old | 1.010 | (0.777) | 0.160 | (0.151) | 0.011 | (0.010) |
| Young | 1.217 | (0.480) | 0.106 | (0.078) | 0.014 | (0.015) |
| Gender | | | | | | |
| Female | 1.046 | (0.758) | 0.147 | (0.150) | 0.008 | (0.010) |
| Male | 1.161 | (0.559) | 0.124 | (0.097) | 0.016 | (0.014) |
| Pre-cocontraction | | | | | | |
| 25% MVC | 1.030 | (0.626) | 0.107 | (0.102) | 0.009 | (0.009) |
| 50% MVC | 1.096 | (0.681) | 0.143 | (0.131) | 0.013 | (0.012) |
| 75% MVC | 1.189 | (0.687) | 0.156 | (0.139) | 0.016 | (0.015) |
| All | 1.105 | (0.663) | 0.135 | (0.126) | 0.012 | (0.013) |

| Shoulder in sagittal plane | *Normalized Stiffness (Nm/rad/kg/m)* $K_1$ | | *Normalized Stiffness (Nm/rad/kg/m)* $K_2$ | | *Normalized Damping (Nms/rad/kg/m)* $B$ | |
|---|---|---|---|---|---|---|
| | Mean | (SD) | Mean | (SD) | Mean | (SD) |
| Age | | | | | | |
| Old | 10.400 | (3.369) | 0.773 | (0.379) | 0.103 | (0.087) |
| Young | 9.321 | (1.774) | 0.806 | (0.470) | 0.090 | (0.071) |
| Gender | | | | | | |
| Female | 11.235 | (3.231) | 0.861 | (0.473) | 0.076 | (0.069) |
| Male | 8.644 | (1.463) | 0.719 | (0.357) | 0.117 | (0.086) |
| Pre-cocontraction | | | | | | |
| 25% MVC | 9.801 | (2.765) | 0.839 | (0.428) | 0.076 | (0.067) |
| 50% MVC | 9.960 | (2.694) | 0.787 | (0.384) | 0.100 | (0.080) |
| 75% MVC | 9.952 | (2.991) | 0.738 | (0.456) | 0.114 | (0.090) |
| All | 9.904 | (2.795) | 0.788 | (0.422) | 0.097 | (0.080) |

**Table 4.2** Results for testing the hypotheses. ANOVA tables for the main effect and the interactions for joint stiffness and damping coefficients at elbow and at shoulder.

| Elbow | $K_1$ | | $K_2$ | | $B$ | |
|---|---|---|---|---|---|---|
| | F | P | F | P | F | P |
| **Main Effect** | | | | | | |
| Age | 5.91 | 0.0207* | 0.01 | 0.9236 | 0.56 | 0.4611 |
| Gender | 9.29 | 0.0045* | 3.98 | 0.0542* | 2.15 | 0.1525 |
| Pre-cocontraction | 26.48 | <.0001* | 4.32 | 0.0216* | 44.08 | <.0001* |
| **Factor Interaction** | | | | | | |
| Age x Gender | 0.44 | 0.5129 | 0.89 | 0.3521 | 0.04 | 0.8485 |
| Age x Pre-cocontraction | 0.15 | 0.8601 | 0.05 | 0.9476 | 9.27 | 0.0006* |
| Gender x Pre-cocontraction | 0.19 | 0.8245 | 0.21 | 0.8129 | 0.21 | 0.8133 |
| Age x Gender x Pre-cocontraction | 0.08 | 0.92 | 0.93 | 0.403 | 0.6 | 0.5525 |

| Shoulder in transverse plane | $K_1$ | | $K_2$ | | $B$ | |
|---|---|---|---|---|---|---|
| | F | P | F | P | F | P |
| **Main Effect** | | | | | | |
| Age | 0.85 | 0.3619 | 1.96 | 0.1713 | 0.16 | 0.6937 |
| Gender | 0.12 | 0.7348 | 0.09 | 0.7699 | 3.94 | 0.0555* |
| Pre-cocontraction | 5.33 | 0.0099* | 6.77 | 0.0034* | 13.18 | <.0001* |
| **Factor Interaction** | | | | | | |
| Age x Gender | 0.42 | 0.5212 | 1.7 | 0.2016 | 0 | 0.9714 |
| Age x Pre-cocontraction | 0.33 | 0.7245 | 0.72 | 0.4923 | 1.06 | 0.3576 |
| Gender x Pre-cocontraction | 0.96 | 0.3945 | 1.8 | 0.1808 | 1.64 | 0.2101 |
| Age x Gender x Pre-cocontraction | 0.48 | 0.6244 | 2.28 | 0.1177 | 0.7 | 0.5046 |

| Shoulder in sagittal plane | $K_1$ | | $K_2$ | | $B$ | |
|---|---|---|---|---|---|---|
| | F | P | F | P | F | P |
| **Main Effect** | | | | | | |
| Age | 0.74 | 0.3972 | 0.16 | 0.6936 | 0.49 | 0.4896 |
| Gender | 8.67 | 0.0059* | 1.22 | 0.2766 | 2.79 | 0.1045 |
| Pre-cocontraction | 1.41 | 0.2585 | 4.77 | 0.0152* | 18.23 | <.0001* |
| **Factor Interaction** | | | | | | |
| Age x Gender | 0.02 | 0.8832 | 0.04 | 0.8403 | 8.61 | 0.006* |
| Age x Pre-cocontraction | 1.48 | 0.2424 | 0.65 | 0.5296 | 0.01 | 0.9922 |
| Gender x Pre-cocontraction | 1.64 | 0.2093 | 0.3 | 0.74 | 2.91 | 0.0683 |
| Age x Gender x Pre-cocontraction | 1.17 | 0.3222 | 5.66 | 0.0077* | 2.87 | 0.0708 |

### 4.3.2 Muscle strength test

Muscle strength varied across all subjects (Figure 4.5). Men were stronger than women and young adults were stronger than old adults in all six tests (see Table 4.3 and appendix). An ANOVA showed a significant difference effect in age and gender ($p < 0.001$).

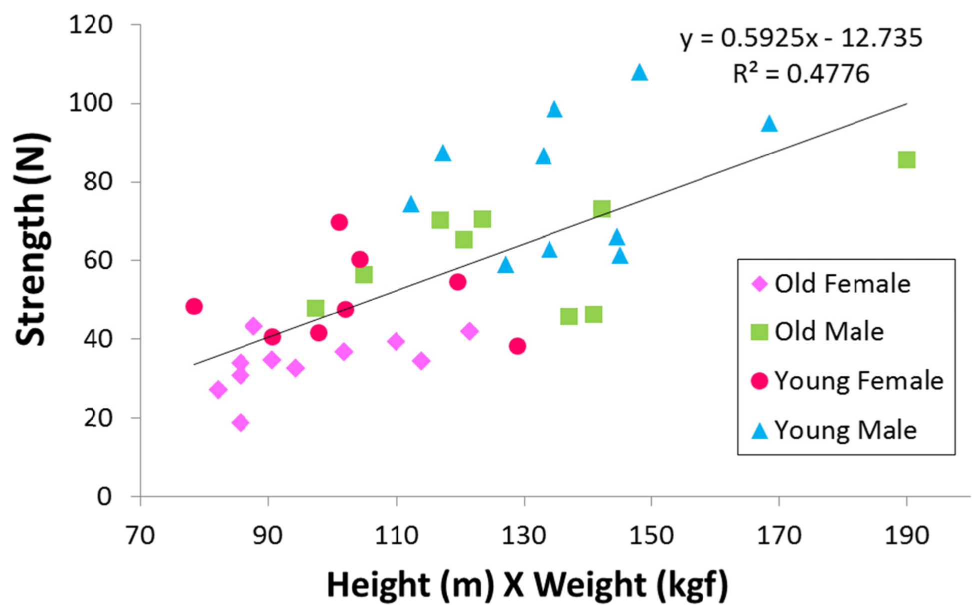



**Figure 4.5** Scatter plot of the shoulder muscle strengths plotted against body size (height times weight) for all subjects. This plot shows the strength of the anterior deltoid muscle (from straight arm pull up (front) test in Appendix Figure 4.a). The other muscle strengths showed similar results.

**Table 4.3** Mean (SD) strength from six different postures (see appendix) for all subjects by age, gender, and age x gender.

| | *Straight arm pull down (side)* | | | | *Straight arm pull up (side)* | | | | *Bending arm pull down* | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Mean | ( | SD | ) | Mean | ( | SD | ) | Mean | ( | SD | ) |
| Age | | | | | | | | | | | | |
| Old | 75.7 | ( | 22.5 | ) | 46.9 | ( | 17.2 | ) | 135.1 | ( | 52.9 | ) |
| Young | 109.8 | ( | 29.8 | ) | 67.8 | ( | 20.9 | ) | 164.1 | ( | 40.3 | ) |
| Gender | | | | | | | | | | | | |
| Female | 71.7 | ( | 20.6 | ) | 42.8 | ( | 12.2 | ) | 118.1 | ( | 38.2 | ) |
| Male | 112.0 | ( | 26.5 | ) | 70.9 | ( | 19.7 | ) | 179.5 | ( | 38.4 | ) |
| Age x Gender | | | | | | | | | | | | |
| Old Female | 60.2 | ( | 13.1 | ) | 36.5 | ( | 7.3 | ) | 102.0 | ( | 36.5 | ) |
| Old Male | 94.7 | ( | 15.7 | ) | 59.7 | ( | 17.4 | ) | 175.6 | ( | 40.2 | ) |
| Young Female | 87.6 | ( | 18.6 | ) | 51.4 | ( | 12.6 | ) | 140.4 | ( | 29.4 | ) |
| Young Male | 127.6 | ( | 24.9 | ) | 81.0 | ( | 16.4 | ) | 183.1 | ( | 38.6 | ) |
| All | 91.9 | ( | 31.1 | ) | 56.8 | ( | 21.6 | ) | 148.8 | ( | 49.0 | ) |

| | *Straight arm pull down (front)* | | | | *Straight arm pull up (front)* | | | | *Bending arm pull up* | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Mean | ( | SD | ) | Mean | ( | SD | ) | Mean | ( | SD | ) |
| Age | | | | | | | | | | | | |
| Old | 81.4 | ( | 27.2 | ) | 46.7 | ( | 17.8 | ) | 108.2 | ( | 39.0 | ) |
| Young | 107.7 | ( | 27.4 | ) | 66.7 | ( | 21.0 | ) | 154.8 | ( | 44.5 | ) |
| Gender | | | | | | | | | | | | |
| Female | 71.4 | ( | 18.6 | ) | 40.8 | ( | 11.8 | ) | 96.0 | ( | 26.8 | ) |
| Male | 116.4 | ( | 21.0 | ) | 71.6 | ( | 17.9 | ) | 164.5 | ( | 37.8 | ) |
| Age x Gender | | | | | | | | | | | | |
| Old Female | 61.0 | ( | 10.8 | ) | 33.9 | ( | 6.9 | ) | 79.7 | ( | 17.3 | ) |
| Old Male | 106.4 | ( | 18.2 | ) | 62.4 | ( | 14.0 | ) | 142.9 | ( | 27.7 | ) |
| Young Female | 85.7 | ( | 17.7 | ) | 50.2 | ( | 10.8 | ) | 118.4 | ( | 20.8 | ) |
| Young Male | 125.4 | ( | 19.9 | ) | 80.0 | ( | 17.5 | ) | 183.9 | ( | 35.9 | ) |
| All | 93.9 | ( | 30.0 | ) | 56.2 | ( | 21.6 | ) | 130.2 | ( | 47.4 | ) |

*Unit : Newtons

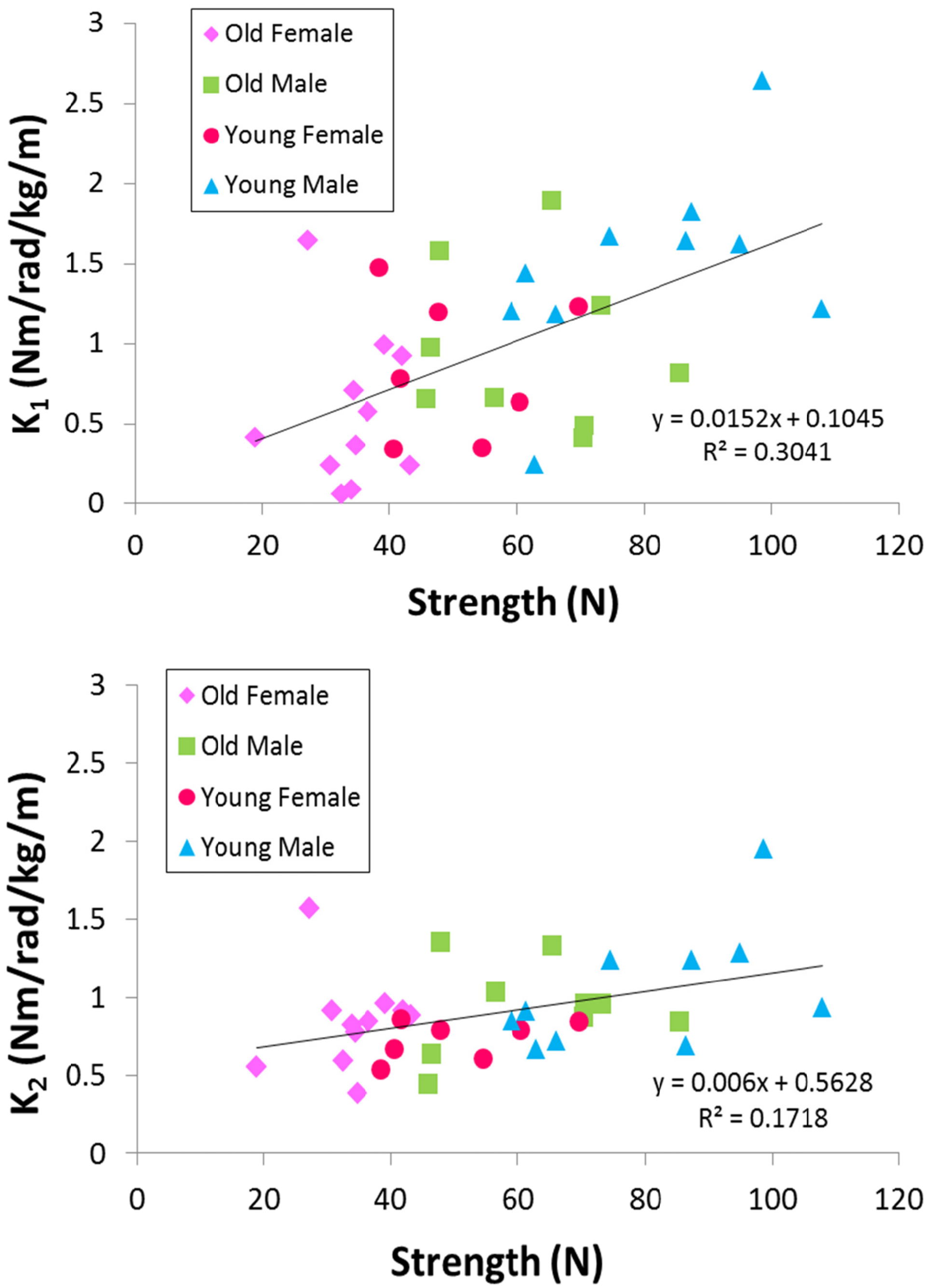
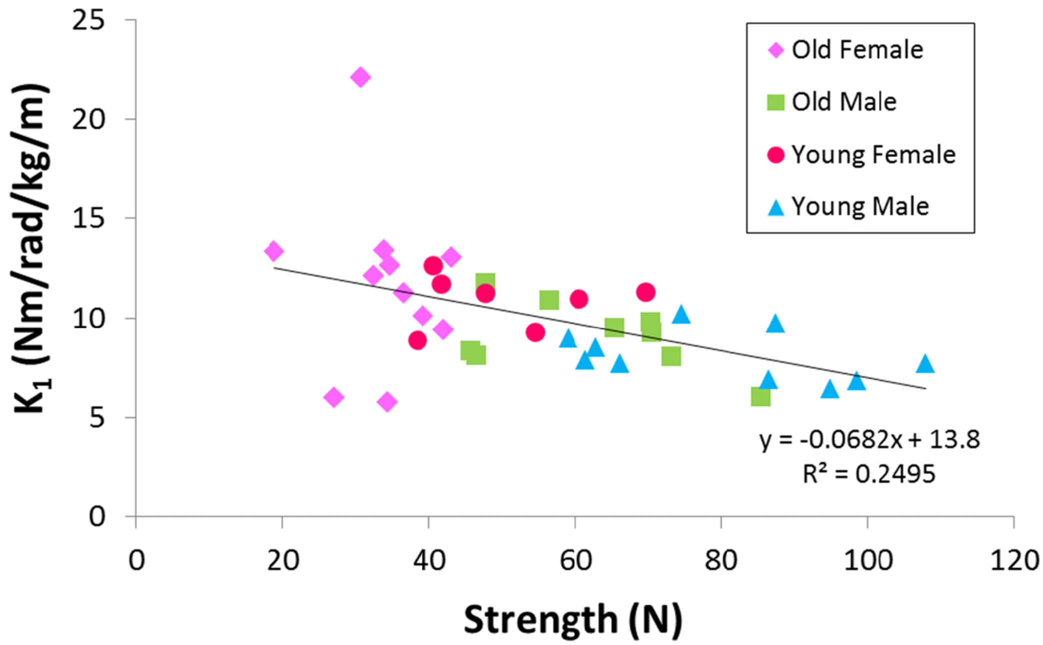



**Figure 4.6** Scatter plot of the elbow normalized stiffness at 75% MVC for all subjects. (Upper: $K_1$, Bottom: $K_2$).

We found a positive relationship between the muscle strength and the normalized stiffness at the elbow (Figure 4.6), but no association between normalized stiffness values and the strength at the shoulder (Figure 4.7 and Figure 4.8).



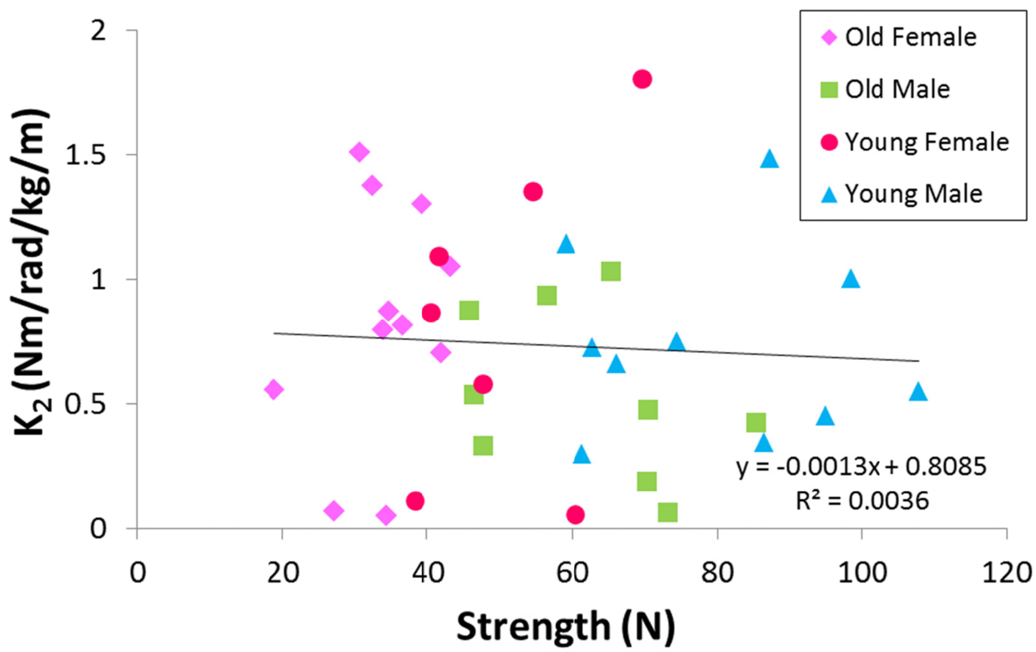
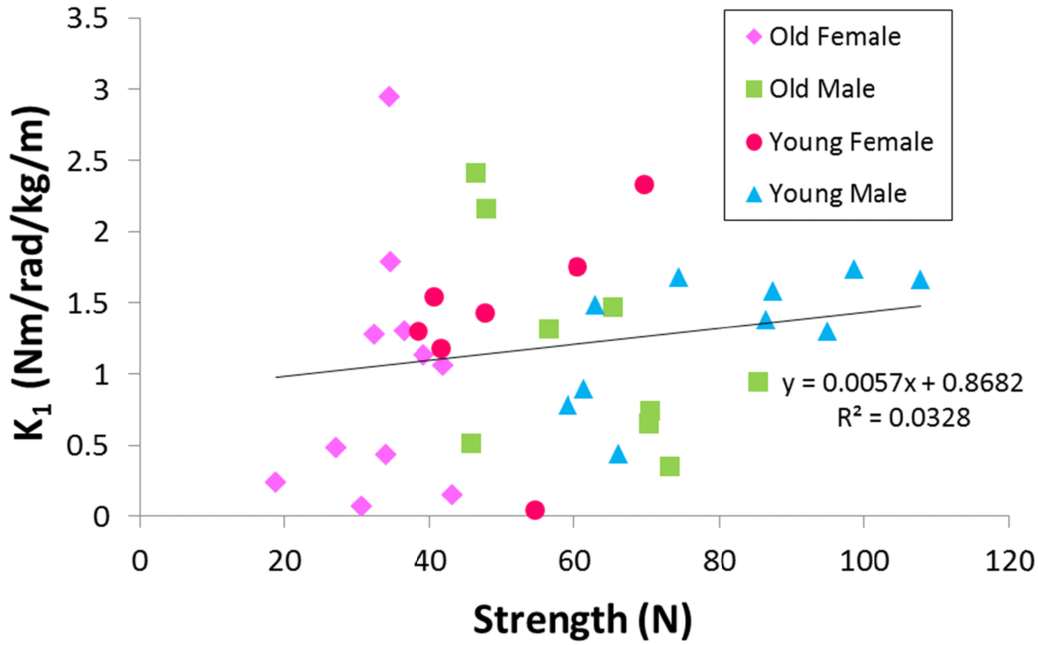




**Figure 4.7** Scatter plot of the normalized shoulder stiffness in sagittal plane 75% MVC. (Upper: $K_1$, Bottom: $K_2$)



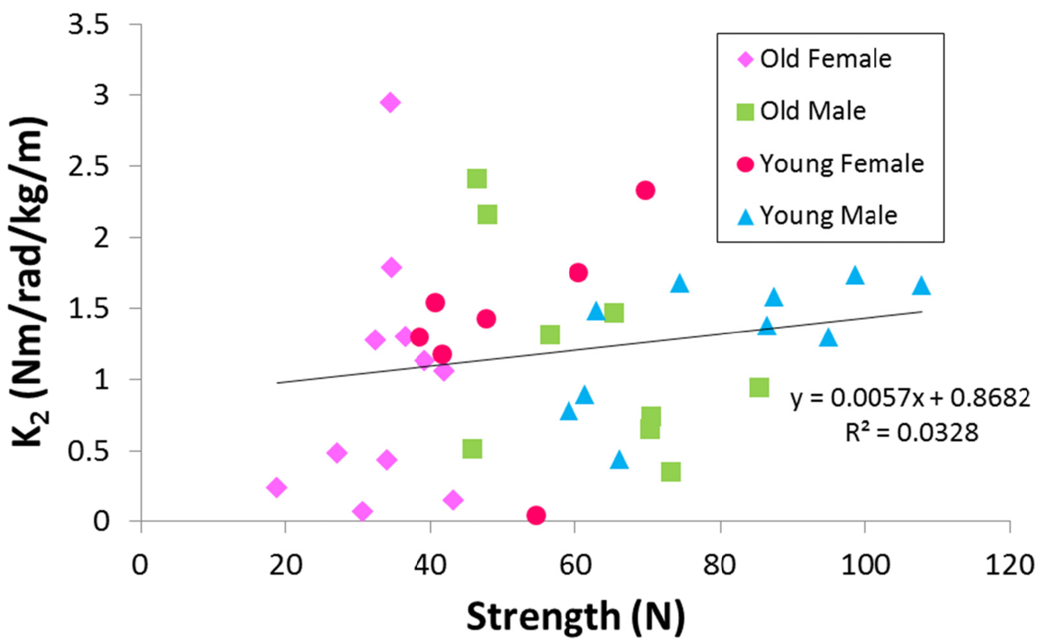




**Figure 4.8** Scatter plot of the normalized shoulder stiffness in transverse plane 75% MVC. (Upper: $K_1$, Bottom: $K_2$)

## 4.4 Discussion

Using a new testing apparatus, we investigated the viscoelastic properties of active upper extremity muscles acting about the elbow and shoulder joints in healthy young and older adults under impulsive arm end-loads. This study provides the first experimental evidence for effect of age, gender and muscle pre-cocontraction level on the rotational stiffness and damping resistance of the elbow and shoulder joints. The main hypotheses were rejected in that the ANOVA (Table 4.2) demonstrated a significant pre-cocontraction level and partially significant age and gender effect on viscoelastic properties. Although a couple of significant interactions were found (for example, age x gender or age x gender x pre-cocontraction in Table 4.2) they do not profoundly affect the main conclusions. The secondary hypothesis was supported in that the strength had a positive relationship with stiffness on elbow extensor however there was no strong association on shoulder.

We found a significant gender effect on the elbow stiffnesses ($K_1$ and $K_2$) as we did in the previous step increasing load study in Chapter 2, but we did not find a significant effect of gender on the shoulder stiffnesses ($K_1$ and $K_2$) values (Table 4.2). However in this study, we found that age affected on the elbow stiffness ($K_1$) value not on the shoulder stiffness values.

In the current study we assumed that the rotational stiffness of the muscles acting about each joint had bilinear properties (Grover et al., 2007; Malamud et al., 1996; Chapter 1 and Chapter 3) in that the initial stiffness ($K_1$) would typically be higher than the later stiffness ($K_2$) under large deflexion. Some $K_1$ and $K_2$ values on elbow in the older adults did not follow this tendency but if we take the mean of $K_1$ and $K_2$ values on elbow then we would find that the absolute value of stiffness was 76.7 Nm/rad for young females and 159.9 Nm/rad for young males (Table 4.4), and the range of non-normalized elbow stiffness and damping coefficients were consistent with previous elbow study in healthy young adults (Chapter 2).

In the current study women exhibited 51% – 80% of men stiffness values and 66% of men damping values and the older adults did 57% − 95% and 76% of young adults stiffness and damping normalized values for elbow joint, respectively. These gender and age differences were also found in the absolute mean values on the elbow in Table 4.4; the older females had the lowest values, next came the young females, the older males had the third highest and the young males had the highest values (Figure 4.6). These results reflect corresponding to age and gender-associated declines in strength and power measurements in the upper extremities in women and men (Metter et al., 1997; Goodpaster et al., 2006; Faulkner et al., 2007). These findings could partially explain why older women are more commonly fail to arrest a fall using their arms, due to their loss in strength and diminution in viscoelastic properties of upper extremity muscles (DeGoede et al., 2003; Lee and Ashton-Miller, 2011).

**Table 4.4** Comparison between current and previous (Chapter 2) results in elbow stiffness and damping properties (non-normalized values).

| *Elbow* | *Current Results* | | | | *Previous Results* | | | |
|---|---|---|---|---|---|---|---|---|
| | *Stiffness (Nm/rad)* *K* | | *Damping (Nms/rad)* *B* | | *Stiffness (Nm/rad)* *K* | | *Damping (Nms/rad)* *B* | |
| | Mean | ( SD ) | Mean | ( SD ) | Mean | ( SD ) | Mean | ( SD ) |
| Old Female | 59.35 | ( 37.15 ) | 2.54 | ( 2.65 ) | | | | |
| Young Female | 76.74 | ( 33.48 ) | 3.22 | ( 3.24 ) | 47.06 | ( 19.28 ) | 1.13 | ( 0.48 ) |
| Old Male | 111.02 | ( 46.89 ) | 4.84 | ( 3.18 ) | | | | |
| Young Male | 159.89 | ( 63.20 ) | 6.83 | ( 4.70 ) | 76.43 | ( 29.10 ) | 2.60 | ( 2.08 ) |
| All | 102.38 | ( 62.17 ) | 4.39 | ( 3.89 ) | 61.75 | ( 24.19 ) | 1.86 | ( 1.28 ) |

*Mean ages 20.3 yrs of nine healthy young females and 25.1 yrs of nine healthy young males. Voluntary moition response measured under a step increase in elbow with 25° of inintial elbow angle. (Lee and Ashton-Miller, 2011)

Although the elderly or women had lower stiffness (including damping) and strength than the young adults or men in the elbow muscles, there is little known about the viscoelastic properties of active shoulder muscles under an impulsive end-loading. This is the why we developed the new apparatus to measure the values of the stiffness and damping coefficient of the shoulder muscles in an active circumstance during impulse. Further we studied the age and gender effects on these viscoelastic properties.

In general, muscle strength decrease with advancing age and female gender (Metter et al., 1997). We also found age and gender differences in strength on elbow and shoulder muscles (Figure 4.5). The values for maximum voluntary elbow and shoulder torques found in the literature are summarized by gender and age in Table 4.5. Mean elbow and shoulder torques in women are similar, however generally shoulder torques are higher than elbow torques among men. This may help explain why we could not find any marked relationship between the stiffness values and the maximum voluntary strength at the shoulder (Figure 4.7 and Figure 4.8).

**Table 4.5** Literature values for joint torque (mean values in Nm) by age, gender, and age x gender.

| | Mean Published Arm Strength (Nm) | | | | | |
|---|---|---|---|---|---|---|
| | 50% Females[a] | 50% Males[a] | Young Females[b] | Young Males[b] | Older Females[b] | Older Males[b] |
| Elbow flexion | 40 | 75 | | | 22[c] | 40[c] |
| Elbow extension | 24 | 46 | | | 25[c] | 39[c] |
| Shoulder flexion | 40 | 92 | 32 | 58 | 16 | 38 |
| Shoulder extension | 33 | 67 | 40 | 72 | 22 | 40 |
| Shoulder adduction | 30 | 67 | | | | |
| Shoulder abduction | 37 | 71 | | | | |

[a]Mean ages 31.3 yr. Isometrically measured strength in elbow and shoulder. (Chaffin and Andersson, 1999; Hagberg et al., 1995; Stobbe, 1982).

[b]Mean ages approximately 25–30 yr for young, 60-80 yr for older adults. Nondominant limb. Isometic strength for shoulder flexors and extensors. (Hughes et al., 1999).

[c]Mean ages approximately 60-90 yr for older adults. Isokinetic strength for elbow flexors and extensors. (Hughes et al., 2001).

We found that the 9.9 Nm/rad/kg/m of the normalized stiffness mean value ($K_1$) and 0.097 Nms/rad/kg/m of the normalized damping mean value (B) of shoulder in sagittal plane (see Table 4.1 and Upper plot in Figure 4.7) were almost 10 times greater than the stiffness and 2.6 times greater than the damping measured at other joints (elbow and shoulder in transverse plane). This result implies that the viscoelastic properties in the flexion and extension muscles at the shoulder should be considered as the important factor for arresting a fall as the ad-/abduction muscles of the shoulder joint.

The present experimental and computational methods include several limitations as we have discussed in Chapter 2. The first limitation was the relatively modest impulsive force generated by the drop-weight-and-rotating beam apparatus which was approximately one fourth of body weight. We have calculated the forces and the torques at the wrist, the elbow and shoulder joints as well as kinematic data. The impulsive force induced the torque of ~25 Nm at elbow and ~50 Nm at shoulder (Figure 4.4) which is less than that would be induced by more than one body weight in a real fall (DeGoede et al., 2003). This was because we did not want to risk injuring any subjects, especially the elderly. Since we did not notify subjects of the exact time of release of weight, they could not anticipate the timing of impact as in a real fall. However, most subjects gave a verbal feedback that this test felt quite similar to a real fall arrest.

A second limitation was that we assumed the proximal end of the clavicle could only rotate, but not translate relative to the torso. A third limitation was that we used the triceps lateral brachii myoelectric muscle activity to indicate the muscle pre-contraction level via biofeedback. Originally we tried to feedback the activity of both muscles (i.e., the triceps for the elbow and the deltoid for the shoulder) but some subjects had difficulty maintaining a constant pre-cocontraction feedback level in both muscles under those conditions so we decided to use only the one muscle for biofeedback.

A fourth limitation is that stiffness and damping values may be overestimated compared to other studies (Frolov et al., 2006 ; Osu et al., 2002; Gomi and Osu, 1998) because the present study was limited to the non-dominant side and we assumed the sagittally-symmetric falls in the computer simulation study (Chapter 3). Most forward fall arrests are not always symmetrical and they probably involve the use of one or other upper extremities rather than both simultaneously. However, we choose to study the one non-dominant side dynamic response for the worst scenario in a real fall.

A fifth limitation is that the stiffness and damping coefficients acting about the elbow were considered independent of those acting about the shoulder. But the portions of both biceps and triceps cross both joints. So these may be some degree of co-dependency between the elbow and shoulder muscle resistance to rotation and that was not studied here. We expect that had we included this effect, the values of $K_1$, $K_2$ and B that we estimated might be more reliable and we could found some significant effects on the shoulder stiffness values.

Despite these limitations, we conclude that the age, gender and especially pre-cocontraction were potent factors to affect the behavior the active viscoelastic properties of upper extremity muscles including the elbow and shoulder joints in healthy young and older adults under large impulsive end-loads.

### 4.5 Acknowledgments

We thank the subjects for their participation and the financial support of PHS Grant P30 AG 024824 is gratefully acknowledged.

## 4.6 Appendix

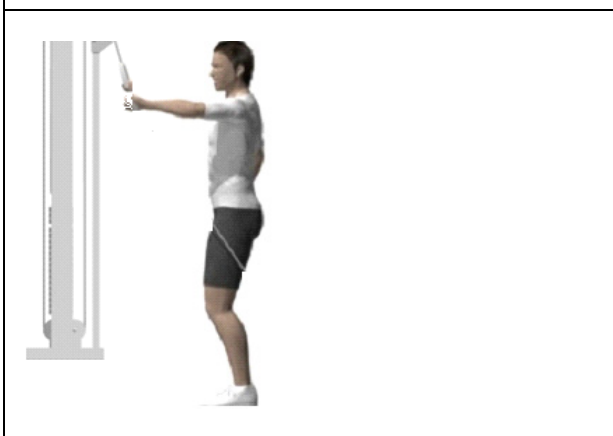
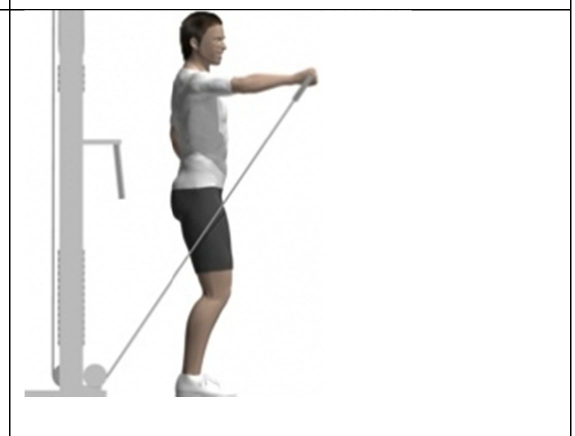
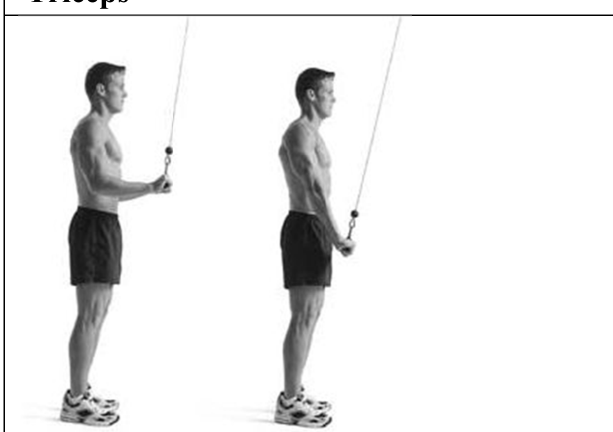
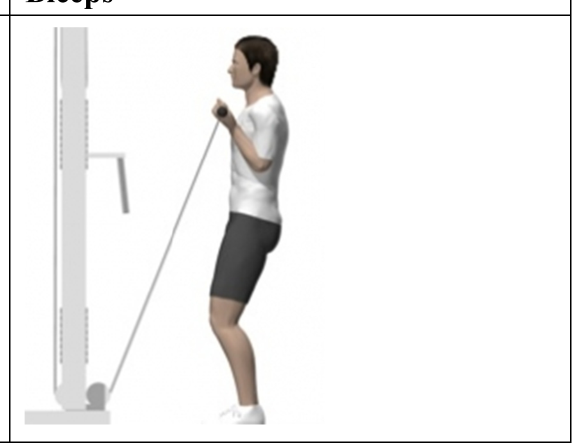
| | |
|---|---|
| a. Straight arm pull down (Side) : **Pectoralis Major** | b. Straight arm pull up (Side) : **Deltoid** |
|  |  |
| c. Straight arm pull down (front) : **Latissimus Dorsi** | d. Straight arm pull up (front): **Anterior Deltoid** |
|  |  |
| e. Bending arm pull down (front) : **Triceps** | f. Bending arm pull up (front) : **Biceps** |
|  |  |

**Figure 4.a** Six different configurations of elbow and shoulder muscle strength test.

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# CHAPTER 5

## Age and Gender Effects on the Proximal Propagation of an Impulsive Force Along the Adult Human Upper Extremity

### 5.1 Abstract

We tested the null hypotheses that neither age, gender nor muscle pre-cocontraction state affect the latencies of changes in upper extremity kinematics or elbow muscle activity following an impulsive force to the hand. Thirty eight healthy young and older adult volunteers lay prone on an apparatus with shoulders flexed 75 degrees and arms slightly flexed. The non-dominant hand was subjected to three trials of impulsive loading with arm muscles precontracted to 25, 50 or 75% of maximum pre-cocontraction levels. Limb kinematic data and upper extremity electromyographic (EMG) activity were acquired. The results showed that pre-cocontraction muscle level ($p < 0.001$) and gender ($p < 0.05$ for wrist and shoulder) affected joint displacement onset times and age affected EMG onset times ($p < 0.05$). The peak applied force ($F_1$) occurred a mean (± SD) 27 (± 2) msec after impact. The latencies for the wrist, elbow and shoulder displacements were 21 ± 3 msec, 29 ± 5 msec and 34 ± 7 msec, respectively. Because the latencies for elbow

flexion and lateral triceps EMG were 23 ± 5 msec and 84 ± 8 msec, respectively, muscle pre-activation rather than stretch reflexes prevent arm buckling under impulsive end loads.

## 5.2 Introduction

The upper extremities are typically the first line of defense to protect the head and torso while bracing for a frontal car crash (Frampton et al., 1997) or arresting a fall to the ground (DeGoede et al., 2003). To prevent a slightly flexed limb from buckling under an impulsive load applied to the hand (DeGoede and Ashton-Miller, 2002; DeGoede et al., 2002; Lee and Ashton-Miller, 2011), the arm muscles are usually precontracted to brace the arm for impact (DeGoede et al., 2003; Lo et al., 2003; Troy and Grabiner, 2007; Lo and Ashton-Miller 2008). Muscle pre-cocontraction may be required because the impulsive ground reaction force on the hand peaks rises so quickly: within a few tens of milliseconds at the hand, and within one hundred milliseconds for the proximal extremity (Hsiao and Robinovitch, 1998; Brown and Loeb, 2000; DeGoede and Ashton-Miller, 2002; DeGoede et al., 2003). Impulsive loads can reach 2–3 times body weight (BW) for a fall from even half standing height and 1–4 kN for a fall from standing height, enough to cause wrist fracture on a hard surface (Frykman, 1967; Dietz et al, 1981; DeGoede and Ashton-Miller, 2002; Lo et al., 2003).

If muscle pre-cocontraction is not used, and one instead has to rely upon neuromuscular reflexes to increase muscular resistance to arm buckling after onset of the impulsive load to the hand, it is not known whether the muscle reflexes are sufficiently rapid that they can effectively increase muscle stiffness before the muscle is forcibly stretched by limb flexion under gravitational and inertial loading. Furthermore, it is not known how long it takes before the elbow begins to flex when an arm is end-loaded. Finally, it is not known whether, once the elbow flexion starts, the triceps muscle reflex is rapid enough to increase muscle resistance to stretch before the elbow flexion phase is complete. One of the goals of this study is to find answers to these questions.

It is known that maximally tensing a striated muscle can increase its stiffness and damping by 50%, so increasing triceps muscle tension to half maximal values should increase its tensile stiffness by about 25% (Blanpied and Smidt, 1993). To voluntarily increase striated muscle force by half-maximal values "as fast as possible" took 74 msec in young females, 87 msec in older females, 92 msec in young males, and 95 msec in older males (Thelen et al., 1996b). So, it takes approximately 90 msec to increase muscle tension by half-maximal values, and muscle stiffness by 25%.

The age and sex of an individual can also affect the time for propagation of the impulsive force along a limb, because they affect the length and mass distribution of the upper extremity. For example, young women typically weigh 17% less than young men and have 8% shorter stature. In addition, young women and men weigh 9% and 5% less than older women and men, respectively (McDowell, 2008). These systematic differences might affect propagation latencies since an impulsive force will take more time to propagate the length of a longer male limb segment than a shorter female limb segment. And larger segmental masses can affect the momentum transfers from one arm segment to the next. For example, Wong et al. found a positive relationship between the longitudinal stress wave velocity and higher mass per unit length of the human bones (Wong et al., 1983). Finally, in terms of reflex latencies, nerve conduction velocity is known to be affected by age; for example, above 60 years the conduction velocity decreases by 1.5% per decade (Norris et al., 1953; Tsuchikane et al., 1995; Cuccurullo, 2004).

The speed of a compressive stress along a human long bone *in vivo* has been reported as 351 $ms^{-1}$ and 266 $ms^{-1}$ for male and female tibia at the medial malleolus (Wong et al., 1995); these value have been corroborated by others on the anteromedial aspect of the tibia (Flynn et al., 2002; Cheng et al., 1995). We might guess that the axial acceleration of the soft tissues (i.e., muscle, fat and skin) overlying a bone would lag that of the bone because of their compliant coupling to the bone (Saha and Lakes, 1977). Hence a compressive stress wave travelling 0.5 m from the hand to the triceps' bony origin will take approximately 1.5 msec. Since, muscle spindles are exquisitely sensitive to stretch (Brown et al., 1967) and vibration (Burke et al., 1976), it is therefore theoretically possible that a muscle spindle in the triceps muscle could begin to sense triceps muscle stretch 2 msec after the onset of the ground reaction force (Norris et al., 1953), even though no elbow flexion has yet occurred. The latency of the monosynaptic reflex is mainly governed by the maximum conduction velocity, 50 $ms^{-1}$, of the largest nerve axons in the nerve to triceps (Taylor, 1984; Norris et al, 1953; Tsuchikane et al., 1995). So, the monosynaptic reflex latency for triceps should take about 20 msec for a 1 m long reflex arc length. This type of non-traditional monosynaptic reflex, which could be initiated by the proximal axial acceleration of the bony origin of the triceps muscle in the direction of the applied end-load to the limb, could potentially be significantly more rapid than the onset of traditional mono- and polysynaptic reflexes related to elbow flexion. The latter have been measured by Dietz et al. (1981) in a fall arrest to be ~25 and ~50 msec, respectively.

The goal of this paper, therefore, was to test the primary (null) hypotheses that neither gender, age, nor level of pre-cocontraction affect the time it takes an impulsive force to propagate proximally along the upper extremity in healthy adults. The secondary hypothesis was that this propagation time is always shorter than the latency of the triceps EMG response to elbow flexion caused by impulsive loading. To help interpret the results, a forward dynamics model was used to explore how the magnitude of hand preload affects impulse propagation times along the upper extremity.

## 5.3 Methods

Ten healthy young males of mean (± SD) age of 25.5 (± 2.7) years, eight healthy young females of 24.5 (± 3.1) years, 9 healthy old males of 69.4 (± 3.4) years and 11 healthy old females of 67.7 (± 2.4) years gave written informed consent to participate in the study, which was approved by the institutional review board. Mean height and mass for the young males were 1.795 (± 0.077) m and 75.88 (± 6.74) kg, for the young females were 1.683 (± 0.060) m and 60.96 (± 7.67) kg, respectively and for the old males were 1.734 (± 0.085) m and 74.73 (± 11.83) kg and for the old females were 1.623 (± 0.039) m and 59.20 (± 7.01) kg, respectively. Subjects were screened by telephone to exclude any chronic illnesses or upper extremity fractures or sprains within the previous year.

An Optotrak Certus camera (Northern Digital, Inc., Waterloo, Canada) was used to measure the displacements of 15 arm, shoulder and neck optoelectronic markers taped to the skin over wrist, elbow and shoulder joint bony landmarks (Figure 5.1). We placed quadruple surface electromyographic (EMG) electrodes (Trigno™ Wireless System, Delsys, Inc., Boston, MA. USA) on the skin over the mid belly of selected arm muscles to measure non-dominant arm muscle activity. Each Trigno$^{TM}$ electrode also contained a triaxial accelerometer from which signals were recorded. Each subject was asked to warm up his/her arm muscles by doing various stretches and several push-ups. We then measured subject's resting and maximum voluntary pre-cocontraction (MVC) EMG levels of the triceps brachii (lateral head) muscles during elbow and shoulder flexion, extension and ab- and adduction by pulling up or pushing down on an handle attached to a vertical cable in series with an uniaxial force transducer (TLL-500, Transducer Techniques, Temecula, CA. USA).

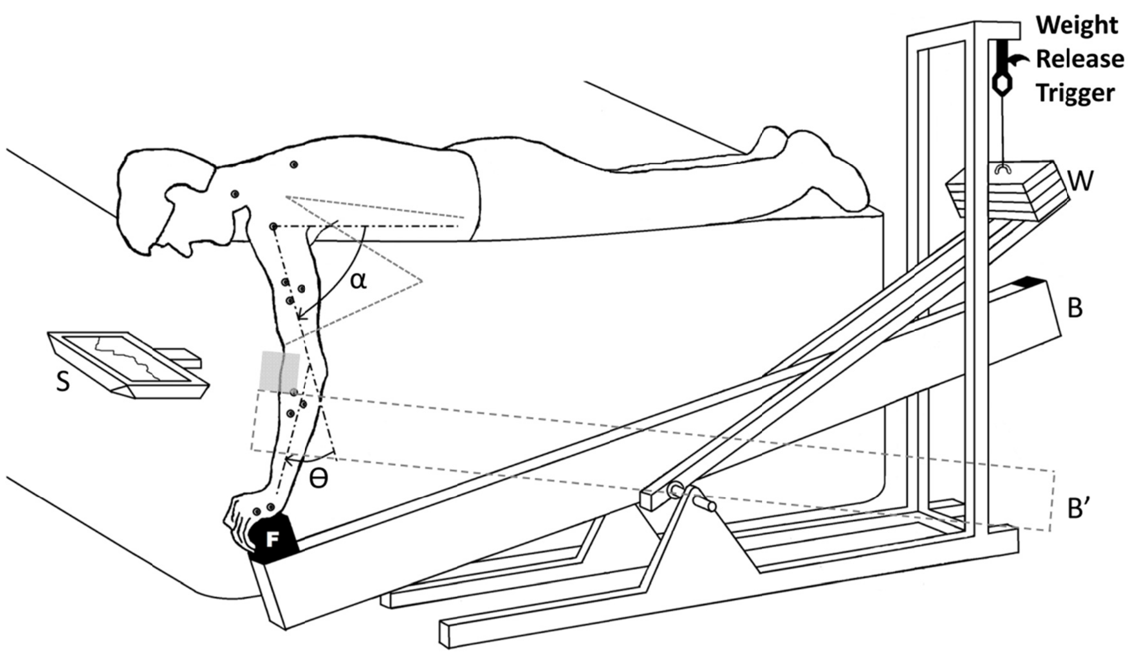



**Figure 5.1** Apparatus for the impulsive end loading test of a human upper extremity. Each subject lay on a padded table with left hand positioned on a force transducer (F). The subject was asked to concentrate on monitoring EMG biofeedback from his/her elbow extensor muscle activity on a display screen (S) and maintaining it at a certain level of effort. A weight (W) of 23 kgf was released onto the end of the lever-arm (B) by a remote trigger, applying an impulsive force to the wrist (at the other end of lever arm) causing elbow flexion and shoulder adduction (the end of the lever-arm position being changed from B to B'). Alpha (α) and theta (Θ) represent shoulder and elbow flexion angles, respectively.

Next, each subject was asked to lie prone on a table with the base of the left hand "lightly touching" a 6 axis force transducer (MC3A-1000, AMTI, Newton, MA. USA) mounted at one end of a 76 mm x 152 mm x 2,032 mm hollow aluminum beam having a rectangular cross-sectional shape and wall thickness 6.35 mm (Figure 5.1). The preload on the transducer was not controlled. The beam was pivoted at its midpoint about a fulcrum formed from a pair of collinear needle bearings and mounted on an axle in the horizontal plane.

The initial flexion angle of the elbow was adjusted to 25 degrees. When ready, the subject was asked to hold his/her hand "lightly" in contact with the force transducer and to concentrate on monitoring EMG biofeedback from his/her lateral head of the triceps muscle activity on a display screen and maintaining triceps activity either at rest, or 25, 50, and 75% MVC values from the main agonist muscle. A weight of 23 kgf was released (see Figure 5.1) from a fixed height of 720 mm to impact the top surface of the other end of the beam, thence applying an upward impulsive force to the wrist via the force transducer, thereby causing elbow flexion, and shoulder adduction and trunk extension. The subject was instructed "not to intervene" before, during and after the weight drop. For example, if the trial was conducted with 75% MVC, the subject was instructed to contract the target muscle steadily at 75% of MVC during the test. At least three trials were conducted at each of the three levels of muscle activation, and these were presented in randomized order.

3-D kinematics data were measured at 280 Hz from 15 infrared-emitting diodes adhered to the skin with double-sided tape. The kinematics and force data were digitally low-pass filtered (MATLAB, The MathWorks, 4th order Butterworth) with cutoff frequencies of 30 Hz and 300 Hz (DeGoede and Ashton-Miller, 2002; Lo et al., 2003). Surface electromyography (EMG) data were collected at 4 kHz. A band-pass $6^{th}$ order filter with breakpoints at 40 Hz and 500 Hz was used to attenuate any movement artefacts in the EMG signal. Then, the EMG data were rectified and digitally low-pass $4^{th}$ order Butterworth filtered with a cutoff frequency of 30 Hz (Thelen et al., 1996a; Dietz et al., 1981). EMG data were normalized by maximum MVC values. The muscle pre-cocontraction state for each trial was determined as the mean value of a 100 msec time window 50 msec before the weight drop.

The onset time of displacement of each joint marker was calculated using MATLAB routines as the intersection of two linearly extrapolated lines: the baseline level line, defined from 100 msec before the impact, and the slope of the signal for 150 msec following the impact (Thelen et al., 1996a). Similarly, the rectified EMG onset time of the target agonist muscle (i.e., lateral triceps brachii) was determined by a MATLAB routine when it exceeded the threshold at the mean plus two standard deviation (SD) value. Peak force values were identified using a MATLAB moving window routine to find local maximum and minimum values.

Typical temporal measurements for each signal in a subject are shown in Figure 5.2. The red circles in the plots (A), (B), (C), and (D) show the onset of each joint marker displacement: $t_a$ was defined as the onset of wrist marker displacement; $t_b$, the onset of elbow rotation ($\Theta_{elbow}$); $t_c$, the onset of elbow linear displacement; and $t_d$, the onset of the shoulder displacement. In the force graph (E), the red circles represent; $t_{0,}$ the onset (0 msec) of the impulsive force ($F_0$); $t_1$, the latency of the first peak in that force signal ($F_1$); $t_2$, the latency of second peak force ($F_2$); $t_3$, the latency of the first minimum

($F_3$) in the force trace between $F_1$ and $F_2$. The bottom plot (F) shows raw (blue dotted line) and the filtered (magenta solid line) EMG data with the onset time indicated by the solid vertical red line, $t_e$.

**Data Analyses**

Descriptive statistics were calculated for peak force, joint marker and EMG onset times. A repeated measures analysis of variance (rm-ANOVA) was used to test the null hypothesis for age, gender, and three different muscle pre-cocontraction levels using SAS 9.3 software. An analysis of covariance (ANCOVA) was used to examine the relationship between preload and pre-cocontraction and their effects on results. A p-value of less than 0.05 was considered statistically significant for the three main effects (primary hypothesis). A Bonferoni correction was used for the interactions.

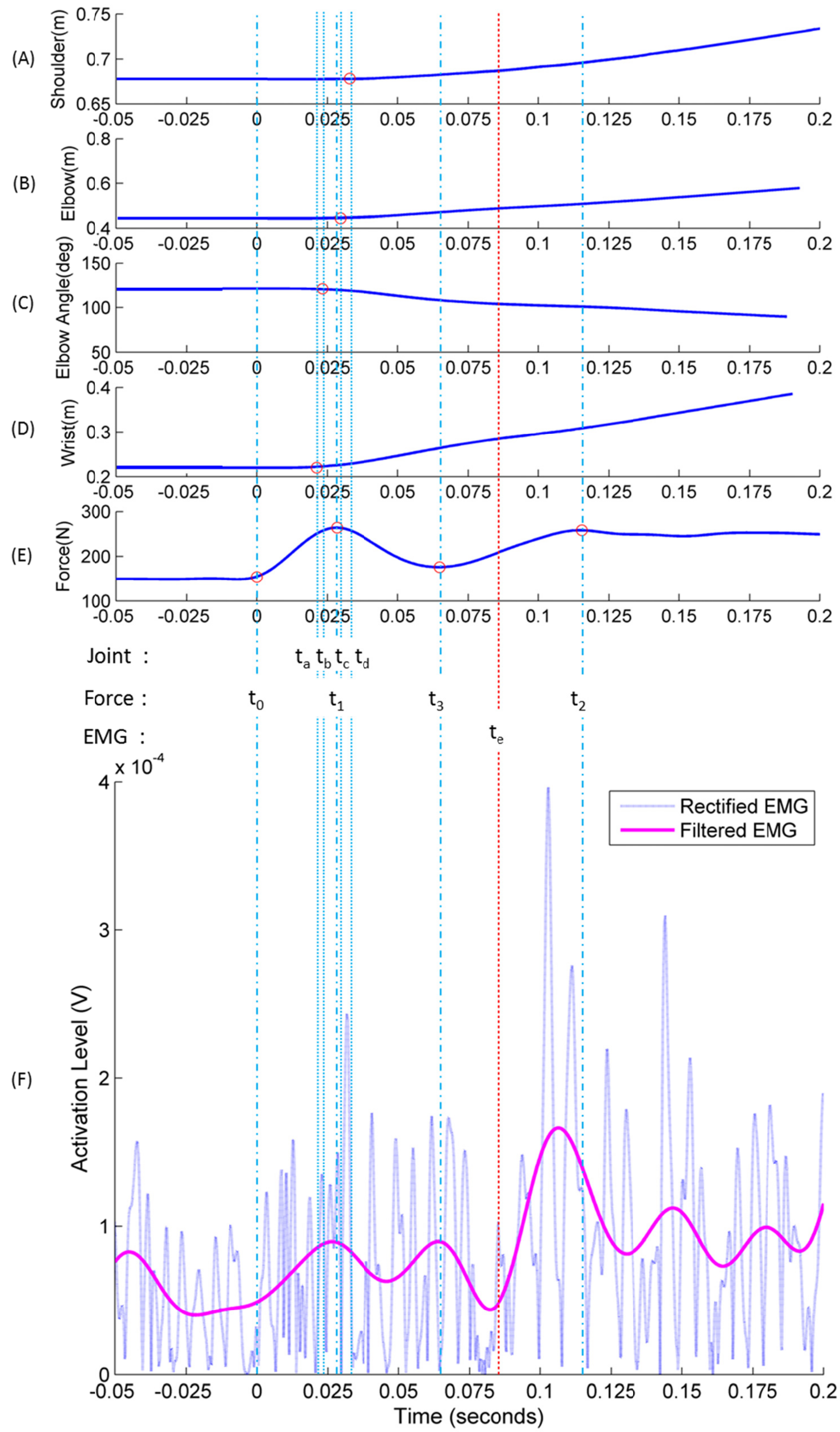
**Forward Dynamics Model**

To examine the effect of hand preload on the results, forward dynamics calculations of upper limb segment kinetics were analyzed using MD Adams$^{TM}$ (MSC. Software Corporation, version 2010). A 3-D, sagittally-symmetric, four-link (including hand, forearm, upper arm and clavicle), lumped parameter, musculoskeletal representation of a 50$^{th}$ percentile male upper extremity was modeled. Arm segments were assumed to be connected by two frictionless revolute joints representing wrist and elbow joints and by two frictionless spherical joints at shoulder and sternoclavicular joints. Segment anthropometric, mass, and inertial properties were assigned based upon the literature (Winter, 2005). The model arm muscles were represented by a torsional spring and damper in parallel at the elbow, and again at the shoulder and sternoclavicular joints, with average tensile stiffness and damping coefficients for either 25%, 50% or 75% MVC levels taken from the literature (Lee and Ashton-Miller, 2011). The model hand-wrist was loaded in a proximal direction with a force time history and peak values equal to the average values measured in each of three blocks of impulsive loading tests. Model kinematic responses were simulated based on preloads of 0, 50 and 100 N, and pre-cocontraction levels of either 25%, 50% and 75% MVC. For each test condition, the mean model-predicted onset times of the linear displacements at each joint and angular displacement at the elbow were calculated to examine the effect of hand preload on impulse transmission times along the extremity.

## 5.4 RESULTS

A typical temporal history for each signal in a subject is shown in Figure 5.2. Across all subjects, the peak applied force ($F_1$) occurred a mean (± SD) 27 (± 2) msec after the onset of the applied force ($t_1$ in Table 5.1). Similarly, the onset times for displacements of the wrist, elbow and shoulder markers were 21 (± 3) msec, 29 (± 5) msec and 34 (± 7) msec, respectively ($t_a$, $t_c$, and $t_d$ in Table 5.1). The corresponding onset times for elbow flexion and lateral triceps brachii muscle activity were 23 (± 5) msec and 84 (± 8) msec, respectively ($t_b$ and $t_e$ in Table 5.1). The second peak ($F_2$) was reached at 120 (± 9) msec and the minimum value ($F_3$) at 63 (± 5) msec, respectively ($t_2$ and $t_3$ in Table 5.1).

In terms of the hypothesis testing, Table 5.2 shows the main effect of age, gender and pre-cocontraction level and the interaction of the onset times of each joint and the latencies of the force signals, and EMG onset time in the lateral triceps brachii. The ANOVA (Table 5.2) demonstrated a significant gender effect, age effect, and pre-cocontraction level effect on onset times. Age affected EMG onset time, gender affected wrist and shoulder displacement onset times, and pre-cocontraction level significantly affected all kinematic onset times, as well as the time to first and second peak applied force. While the higher pre-cocontraction level was associated with a more rapid onset time for each joint marker, the lower pre-cocontraction level was associated with a shorter time to reach the first peak force ($F_1$). Although the wrist marker in males started to displace earlier, the latencies of propagation proximal to the shoulder marker was longer than in the females. Young adults had significantly shorter EMG onset times than older adults ($p < 0.05$).



(A)
Shoulder(m)
0.75
0.7
0.65
(B)
Elbow(m)
0.8
0.6
0.4
(C)
Elbow Angle(deg)
150
100
50
(D)
Wrist(m)
0.4
0.3
0.2
(E)
Force(N)
300
200
100
-0.05 -0.025 0 0.025 0.05 0.075 0.1 0.125 0.15 0.175 0.2
Joint :
$t_a$ $t_b$ $t_c$ $t_d$
Force :
$t_0$ $t_1$ $t_3$ $t_2$
EMG :
$t_e$
(F)
x $10^{-4}$
Activation Level (V)
4
3
2
1
0
Rectified EMG
Filtered EMG
Time (seconds)

**Figure 5.2** Temporal plots for one trial in Subject OFZC showing the onset times of landmark movement (A-D), measured impulsive force (E), and EMG in the lateral triceps brachii (F). $t_0$ represents the onset (0 msec) of the impulsive force ($F_0$); $t_1$, the latency of the first peak in that force signal ($F_1$, 27 msec); t2, the latency of second peak force ($F_2$, 120 msec); $t_3$, the latency of the first minimum ($F_3$, 63 msec) in the force trace between $F_1$ and $F_2$; ta, the onset of wrist marker displacement (21 msec); $t_b$, the onset of elbow rotation ($\Theta_{elbow}$, 23 msec); $t_c$, the onset of elbow linear displacement (29 msec); and $t_d$, the onset of the shoulder displacement (34 msec). The bottom plot (F) shows raw (blue dotted line) and the filtered (magenta solid line) EMG data with the onset time indicated by the solid vertical red line, $t_e$ (84 msec).

Average values of the initial force ($F_0$), the first peak force ($F_1$) and the second peak force ($F_2$) were higher in young and male subjects, as well as in higher pre-cocontraction level condition, than in old or female subjects, and in the lower pre-cocontraction level condition (Table 5.3). The muscle pre-cocontraction level was found to have a significant effect on the onset times of arm joint displacements in the current study. The relationship between the pre-cocontraction level and the preload, which was represented as $F_0$ (the initial force), is shown in Tables 5.3 and 5.4. The other two peak forces ($F_1$, $F_2$) including $F_0$ were significantly affected by the pre-cocontraction level ($p < 0.0001$).

**Table 5.1** Mean (SD) onset times of each joint and the latencies of the force signals, and EMG onset time in the lateral triceps brachii by age, gender, and level of pre-cocontraction. Please see text for definitions of the times.

| | Wrist onset time $t_a$ | | Elbow angle onset time $t_b$ | | $F_1$ (first peak) time $t_1$ | | Elbow onset time $t_c$ | |
|---|---|---|---|---|---|---|---|---|
| | Mean | (SD) | Mean | (SD) | Mean | (SD) | Mean | (SD) |
| Age | | | | | | | | |
| Old | 21.3 | (2.7) | 23.9 | (3.7) | 27.2 | (2.2) | 27.6 | (4.5) |
| Young | 20.9 | (3.4) | 21.7 | (5.2) | 27.4 | (2.3) | 29.7 | (4.8) |
| Gender | | | | | | | | |
| Female | 22.2 | (3.0) | 23.3 | (5.4) | 26.5 | (2.1) | 28.4 | (4.2) |
| Male | 20.0 | (2.7) | 22.5 | (3.7) | 28.1 | (2.1) | 28.8 | (5.3) |
| Pre-cocontraction | | | | | | | | |
| 25% | 22.0 | (2.7) | 23.5 | (4.5) | 26.4 | (1.7) | 30.2 | (5.2) |
| 50% | 20.9 | (3.2) | 22.6 | (4.6) | 27.5 | (2.1) | 27.8 | (4.9) |
| 75% | 20.5 | (3.1) | 22.6 | (4.8) | 27.9 | (2.6) | 27.7 | (3.6) |
| All | 21.1 | (3.0) | 22.9 | (4.6) | 27.3 | (2.2) | 28.6 | (4.7) |

| | Shoulder onset time $t_d$ | | $F_3$ (minimum) time $t_3$ | | EMG onset time $t_e$ | | $F_2$ (second peak) time $t_2$ | |
|---|---|---|---|---|---|---|---|---|
| | Mean | (SD) | Mean | (SD) | Mean | (SD) | Mean | (SD) |
| Age | | | | | | | | |
| Old | 32.6 | (5.8) | 63.6 | (3.3) | 86.6 | (7.8) | 123.5 | (7.7) |
| Young | 36.4 | (7.3) | 61.4 | (5.7) | 80.7 | (8.5) | 115.7 | (7.4) |
| Gender | | | | | | | | |
| Female | 32.1 | (5.4) | 63.6 | (4.1) | 84.0 | (9.5) | 123.3 | (8.0) |
| Male | 36.7 | (7.3) | 61.5 | (5.1) | 83.5 | (7.3) | 116.3 | (7.5) |
| Pre-cocontraction | | | | | | | | |
| 25% | 35.8 | (6.8) | 61.8 | (4.8) | 84.3 | (7.7) | 121.9 | (8.7) |
| 50% | 34.5 | (6.7) | 62.6 | (4.5) | 85.5 | (9.9) | 119.2 | (8.0) |
| 75% | 33.0 | (6.9) | 63.3 | (4.9) | 81.5 | (7.8) | 118.3 | (8.5) |
| All | 34.4 | (6.8) | 62.6 | (4.7) | 83.8 | (8.6) | 119.8 | (8.5) |

**Table 5.2** Results for testing the hypotheses. ANOVA tables for the main effect and the interactions affecting the displacement onset times at each joint, the latencies of the force signals, and the EMG onset time for the lateral triceps brachii. In this and Table 5.4, the values of the F-statistic (F) and probability (P) are given for each variable.

| | Wrist onset time $t_a$ | | Elbow angle onset time $t_b$ | | $F_1$ (first peak) time $t_1$ | | Elbow onset time $t_c$ | |
|---|---|---|---|---|---|---|---|---|
| | *F* | *p* | *F* | *p* | *F* | *p* | *F* | *p* |
| Main effect | | | | | | | | |
| Age | 0.05 | 0.8185 | 1.93 | 0.1736 | 0.03 | 0.8727 | 2.46 | 0.1264 |
| Gender | 7.33 | 0.0105* | 0.19 | 0.6652 | 6.96 | 0.0125* | 0.08 | 0.7828 |
| Pre-cocontraction | 16.08 | <0.0001* | 12.13 | 0.0001* | 9.67 | 0.0005* | 15.76 | <0.0001* |
| Factor interaction | | | | | | | | |
| Age × gender | 1.28 | 0.2661 | 0.6 | 0.4448 | 0.2 | 0.6559 | 4.85 | 0.0345* |
| Age × pre-cocontraction | 16.62 | <0.0001* | 1.17 | 0.3217 | 0.49 | 0.6178 | 4.27 | 0.0222* |
| Gender × pre-cocontraction | 4.11 | 0.0252* | 3.79 | 0.0328* | 0.89 | 0.4188 | 0.04 | 0.9618 |
| Age × gender × pre-cocontraction | 0.4 | 0.6719 | 1.12 | 0.3384 | 0.54 | 0.5883 | 0.47 | 0.6279 |

| | Shoulder onset time $t_d$ | | $F_3$ (minimum) time $t_3$ | | EMG onset time $t_e$ | | $F_2$ (second peak) time $t_2$ | |
|---|---|---|---|---|---|---|---|---|
| | *F* | *p* | *F* | *p* | *F* | *p* | *F* | *p* |
| Main effect | | | | | | | | |
| Age | 3.74 | 0.0614 | 2.22 | 0.1451 | 10.04 | 0.0032* | 12.53 | 0.0012* |
| Gender | 6.21 | 0.0177* | 2.24 | 0.1436 | 0.01 | 0.92 | 9.38 | 0.0043* |
| Pre-cocontraction | 8.61 | 0.0009* | 2.71 | 0.0809 | 2 | 0.1504 | 9.67 | 0.0005* |
| Factor interaction | | | | | | | | |
| Age × gender | 4.25 | 0.047* | 2.43 | 0.1282 | 0.22 | 0.6428 | 0.47 | 0.4997 |
| Age × pre-cocontraction | 0.52 | 0.5994 | 0.66 | 0.5235 | 0.14 | 0.8665 | 1.22 | 0.3086 |
| Gender × pre-cocontraction | 2.88 | 0.0697 | 0.5 | 0.6084 | 0.74 | 0.4851 | 1.64 | 0.2086 |
| Age × gender × pre-cocontraction | 1.63 | 0.2111 | 0.19 | 0.8284 | 0.11 | 0.8973 | 0.39 | 0.682 |

Since subjects had some difficulty in maintaining steady cocontraction levels prior to loading, we analyzed the mean (± SD) RMS EMG activity at the 25, 50 and 75% cocontraction levels over the last 100 msec prior to impact. These values were 0.055 (± 0.015) mV, 0.077 (± 0.021) mV and 0.101 (± 0.035) mV, respectively.

The increase in elbow flexion angle in the first 150 msec following impact averaged (± SD) 21.1 (± 5.4) and 27.1 (± 6.4) degrees in the young and old, respectively; 20.1 (± 4.0) and 28.1 (± 6.4) degrees in the males and females, respectively; and 25.1 (± 6.1), 24.1 (± 6.8) and 23.5 (± 7.0) for the 25, 50 and 75% MVC trials, respectively.

The forward dynamic upper extremity model predicted that increasing muscle stiffness had less of an effect on joint displacement onset times than increasing the preload on the hand. For example, the model joint displacement onset times ($t_a - t_d$) were shortened by 5% when the pre-cocontraction level was increased from 25 to 75%MVC, but shortened by an average of 15% when the preload was increased from 0 to 100 N (Table 5.5). Pre-cocontraction was predicted to have a larger effect on the increase in elbow angle than preload in this model.

**Table 5.3** Means (SD) values for the magnitudes of the preload force ($F_0$) and two peak forces ($F_1$, $F_2$) by age, gender and pre-cocontraction level.

| | $F_0$ | | $F_1$ | | $F_2$ | |
|---|---|---|---|---|---|---|
| | Mean | (SD) | Mean | (SD) | Mean | (SD) |
| Age | | | | | | |
| Old | 90.3 | (54.7) | 237.4 | (85.3) | 222.1 | (80.6) |
| Young | 114.9 | (69.6) | 275.4 | (77.7) | 277.2 | (73.1) |
| Gender | | | | | | |
| Female | 78.9 | (49.3) | 214.3 | (62.2) | 192.2 | (53.1) |
| Male | 125.1 | (67.2) | 296.4 | (82.5) | 304.2 | (65.1) |
| Pre-cocontraction | | | | | | |
| 25% | 78.6 | (45.3) | 226.9 | (71.1) | 224.8 | (70.6) |
| 50% | 106.2 | (63.1) | 261.7 | (83.4) | 253.5 | (81.7) |
| 75% | 121.0 | (71.9) | 277.5 | (89.2) | 266.2 | (88.1) |
| All | 102.0 | (63.1) | 255.4 | (83.6) | 248.2 | (81.6) |

Unit: Newtons.

**Table 5.4** ANOVA table for testing the effects of factors affecting the magnitudes of preload force (**$F_0$**) and the two peak forces ($F_1$, $F_2$).

| | Preload force ($F_0$) | | First peak force ($F_1$) | | Second peak force ($F_2$) | |
|---|---|---|---|---|---|---|
| | $F$ | $p$ | $F$ | $p$ | $F$ | $p$ |
| Main effect | | | | | | |
| Age | 1.44 | 0.2385 | 1.92 | 0.1749 | 6.98 | 0.0124* |
| Gender | 6.27 | 0.0173* | 12.96 | 0.001* | 40.79 | <0.0001* |
| Pre-cocontraction | 28.96 | <0.0001* | 30.55 | <0.0001* | 34.39 | <0.0001* |
| Factor interaction | | | | | | |
| Age × gender | 2.5 | 0.1228 | 3.29 | 0.0786 | 1.37 | 0.2495 |
| Age × pre-cocontraction | 2.3 | 0.1154 | 2.56 | 0.0923 | 3.06 | 0.06 |
| Gender × pre-cocontraction | 0.81 | 0.4544 | 1.32 | 0.2797 | 2.9 | 0.0685 |
| Age × gender × pre-cocontraction | 0.88 | 0.4223 | 0.62 | 0.5416 | 0.02 | 0.9756 |

* $p < 0.05$.

**Table 5.5** The forward computer model simulation results showing how the displacement onset times at each joint are predicted to be affected by both the preload and the pre-cocontraction conditions. Their effects on the predicted increase in elbow flexion angle that occurred over the first 150 msec following impact are shown in the last column.

| | Wrist onset time $t_a$ (ms) | | | Elbow angle onset time $t_b$ (ms) | | | Elbow onset time $t_c$ (ms) | | | Shoulder onset time $t_d$ (ms) | | | Elbow flexion angle $\Delta\theta_E$ (°) | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Pre-cocontraction | | | Pre-cocontraction | | | Pre-cocontraction | | | Pre-cocontraction | | | Pre-cocontraction | | |
| Preload | 25% MVC | 50% MVC | 75% MVC | 25% MVC | 50% MVC | 75% MVC | 25% MVC | 50% MVC | 75% MVC | 25% MVC | 50% MVC | 75% MVC | 25% MVC | 50% MVC | 75% MVC |
| No (0 N) | 28.1 | 27.9 | 27.1 | 28.8 | 28.1 | 27.6 | 35.1 | 34.3 | 33.6 | 37.9 | 37.3 | 36.4 | 40.4 | 29.8 | 22.6 |
| Low (50 N) | 26.9 | 25.4 | 25.1 | 26.5 | 25.9 | 25.4 | 32.8 | 32.1 | 31.3 | 35.8 | 35.1 | 34.3 | 41.3 | 30.5 | 23.1 |
| High (100 N) | 24.2 | 23.2 | 22.4 | 24.9 | 23.3 | 22.8 | 30.2 | 29.4 | 28.6 | 33.3 | 32.6 | 31.8 | 42.2 | 31.1 | 23.7 |

## 5.5 Discussion

This study provides the first experimental evidence for the effect of age, sex and muscle pre-cocontraction level on the kinematics of the upper extremity loaded under large impulsive end-loads. The main hypotheses were rejected in that the ANOVA (Table 5.2) demonstrated a significant gender effect, age effect, and pre-cocontraction level effect on onset times. The secondary hypothesis was supported in that even the longest latency to onset of the shoulder marker in any subject was approximately 40 msec, with the onsets of displacements at the wrist and elbow being shorter (Table 5.1), and all these latencies being substantially shorter than the mean EMG onset time of the lateral triceps brachii measured at 84 msec (Figure 5.3).

Although several significant interactions were found (i.e., age x gender, age x pre-cocontraction or gender x pre-cocontraction, Table 5.2) they do not seriously affect the main conclusions that gender, age and precontraction level affect onset times.

As far as the secondary hypothesis is concerned, the results suggest that the elbow began to flex 6 msec before the origin of the triceps started to displace, so the triceps EMG response at 84 msec was most likely triggered by arm flexion rather than axial movement of the bony origin of the triceps muscle in the direction of the impulsive force, as discussed in the Introduction. The results also suggest that there is not sufficient time for longer loop neural reflexes to modulate arm buckling resistance before the 84 msec onset of triceps stretch. This latter was either caused by induced elbow flexion (onset of 23 msec) or onset of the proximal displacement of the origin of the triceps muscle near the acromion marker (34 msec). Given the ~90 msec Thelen et al.(1996a) found it takes to increase tension, and therefore tensile stiffness, by half-maximal values (see 5.2

Introduction), it is highly unlikely that any muscle reflex can significantly increase the tensile stiffness of the triceps before it is forcibly stretched at 23 msec. This is true even if the onset of the impulsive force was so rapid that it triggered a flexural stress wave along the bones of the upper arm to set up longitudinal vibrations that initiated a stretch reflex in the triceps muscle.

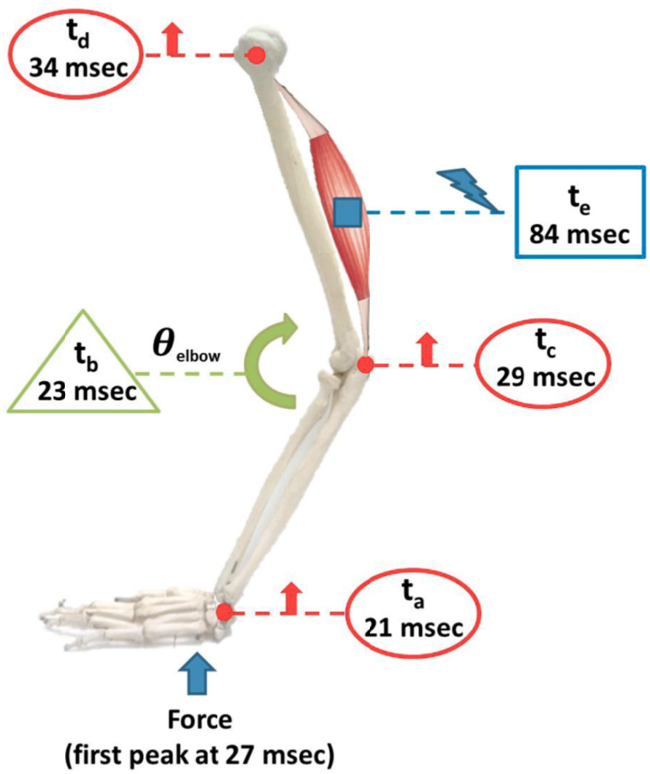



**Figure 5.3** Illustration of mean onset times of landmark movement and EMG onset time across all subjects. The red circles indicate the onset of linear displacements at the wrist, elbow and shoulder joints. The green triangle represents the onset of rotation at the elbow joint. The blue rectangle denotes onset of lateral triceps brachii EMG signal. Measures of data variability are provided in Table 5.1.

The 84 msec onset of lateral triceps brachii muscle activity after impact is consistent with empirical studies conducted over the past 60 years (for review, see

Pruszynski and Scott, 2012). For example, typical elbow muscle responses following stretch yield an initial peak reflex response 20–50 msec after the perturbation (the 'short-latency response') based on a monosynaptic contribution (Pierrot-Deseilligny and Burke, 2005). A second peak, occurring 50–100 msec after the perturbation, is termed the 'long-latency stretch response' based on polysynaptic contributions. Our 84 msec triceps response therefore belongs in this latter category. The final myoelectric burst, called the 'voluntary response', occurs after 100 msec and involves feedback control processes ('relatively slow motor response') (Kurtzer et al., 2010). These were not considered in the present study.

The forward dynamic simulation model predictions (Table 5.5) suggested that the preload applied by the hand to the force transducer before it started to accelerate likely affects the displacement onset times at each joint. The zero preload condition (0 N) approximated the *in vivo* situations of arresting a fall to the ground (DeGoede et al., 2003), while larger preloads represent bracing for a frontal car crash (Frampton et al., 1997). The forward dynamic model predictions were verified by comparing the predicted joint displacement onset times for the range of hand preload conditions with the experimental results (c.f., Tables 5.1 and 5.5). The experimental results showed that the preload ($F_0$) was significantly affected by the pre-cocontraction level and gender (Table 5.4) so one cannot separate the effects of the higher pre-cocontraction levels from those caused by the larger preloads. But the forward dynamics model allows one to separate those effects (Table 5.5). In addition, the ANCOVA results showed that the effect of a preload on arm kinematics during the 25% pre-cocontraction level was significantly greater than that at 50% and 70% levels.

In general, the higher the arm muscle pre-cocontraction level, the faster the stress wave can travel through the stiffer muscles overlying the bone, and this explains the shorter onset time of the wrist, elbow and shoulder joint displacements at higher pre-cocontraction levels (Saha and Lakes, 1977). A confounder was that the higher pre-cocontraction levels were also associated with high preloads on the stationary hand/wrist of a subject. However in our experiment, the presence of that preload was a safety measure to prevent the beam and force transducer accelerating so as to cause an excessive, and therefore potentially unsafe, impact force on the stationary hand/wrist of an older subject (Augat et al., 1998).

Limitations of our methods include the possible presence of movement artefact in the triceps muscle activity responses. We noticed a large positive cross-correlation coefficient, ranging from 0.68 to 0.98, between the EMG signal and the measured linear accelerations of the wireless EMG electrode/amplifier over the first 84 msec post-impact. The linear accelerations were measured from the onboard 3-axis Trigno™ accelerometers during the 40 msec post-impact time frame (Figure 5.4). The presence of motion artefact on the Trigno™ wireless system is surprising given the double differential design of the preamplifier which should have attenuated signals common to the two pairs of electrodes. We cannot exclude the possibility that a monosynaptic stretch reflex might have increased EMG signal in the 25 msec after impact, but given that a maximum EMG pre-cocontraction takes 88 msec (Thelen et al., 1996a) to peak and that it takes 90 msec for muscle to reach 50% MVC (Thelen et al., 1996b), it is unlikely that monosynaptic reflexes could have had much mechanical effect prior to 80 msec.

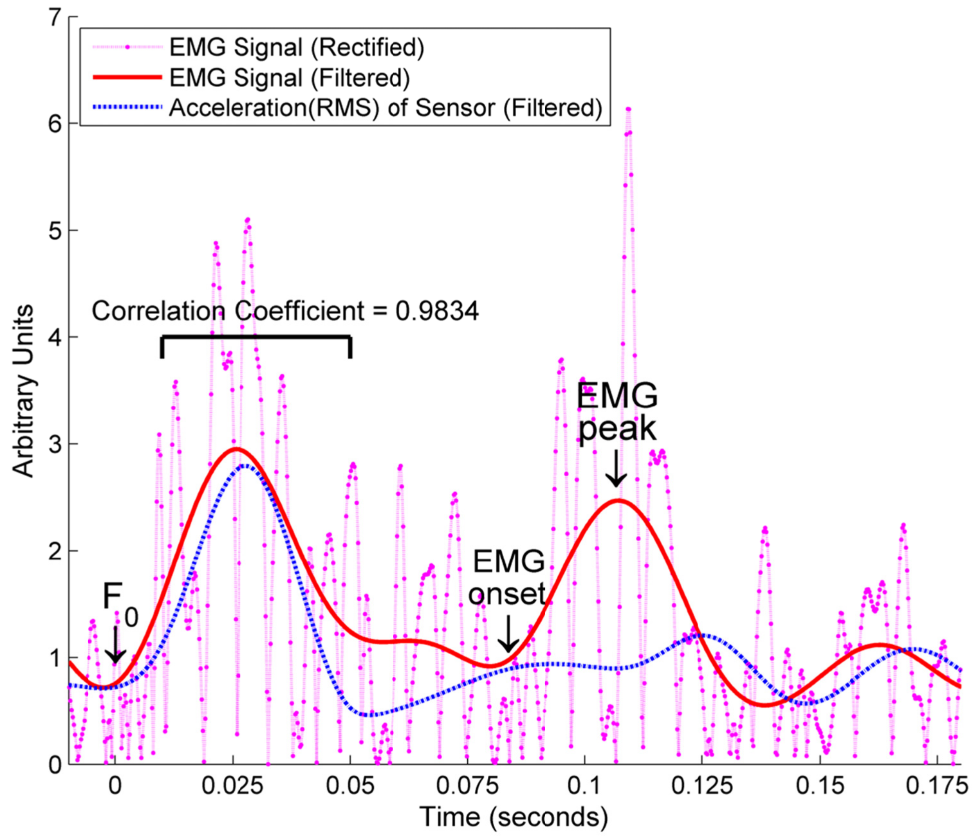



**Figure 5.4** Temporal plot for one trial in Subject OMYB showing the EMG signal (in red line) of the wireless EMG sensor and the acceleration (RMS in blue dashed line) of the 3-axis accelerometer built in to the wireless EMG sensor. Correlation coefficient was calculated during the interval from 10 to 50 msec after the impact ($F_0$).

A second limitation was the fact that some subjects had difficulty using biofeedback to maintain a constant pre-cocontraction feedback level. This likely increased the variability observed in the pre-impact triceps pre-cocontraction level.

A third limitation was the lack of data on the onset of shoulder extension or abduction rotation which would have informed the question of whether shoulder flexor and/or adductor muscle reflexes are rapid enough to change shoulder muscle stiffness before those muscles are stretched. However, given the onset of shoulder motion at 34 msec, the present results suggest this is unlikely, even if the muscle reflex latencies for triceps have shorter reflex arcs.

A fourth limitation related to the relatively modest impulsive force induced by the drop-weight-and-rotating beam apparatus which amounted to approximately 0.25* BW (Table 5.3). While it sufficed to impart a realistic “jolt” to accelerate both arm segments and torso upwards by approximately 10 cm, it is certainly less than the more than 1*BW force that occurs during falls onto a hard surface (see review in Introduction). However, subject safety was paramount in these experiments involving older adults, especially since we did not inform subjects of the exact time of release; they could not anticipate the timing of the impact as in a real fall (Lee and Ashton-Miller, 2011; Troy and Grabiner, 2007; DeGoede and Ashton-Miller, 2002; DeGoede et al., 2002). This means that the neural ‘set’ and reticular activation of the motor control system might have been less in these subjects than during a real fall arrest. For example, the vestibular system was inactivated by the stimulus, there was no visual stimulus from an on-rushing surface to clue one in to when impact would occur, and there was imprecise anticipation of when ‘ground contact’ would occur.

Our experiment helps one consider upper extremity responses in the act of bracing oneself against steering wheel or the dashboard of a car before a frontal crash whose exact timing is uncertain. Our results on the latencies of the kinematic movements are conservative since in real falls Dietz et al. (1981) have shown latencies half (i.e., 10 – 20 msec) those measured in this experiment (i.e., 27 msec, Figure 5.2 E). Thus the longer latency to a lower peak loading we used in this paper is more reflective of landing on or striking a soft rather than a hard surface (Cummings and Nevitt, 1994). Hence, in a real fall, there would be even less time for reflexes to increase muscle forces than in the present experiment.

The most important result from this study is the marked effect of pre-cocontraction level and gender on the propagation of an impulse along the upper extremity. Furthermore, the "preflex", resulting from forced stretch of pre-activated muscle, with its increased stiffness and viscosity property states, is more important than reflexes in dictating the response of an arm to an impulsive end load (Brown and Loeb, 2000). We conclude that there is insufficient time for muscle reflexes to significantly increase muscle resistance to arm buckling when using the arm to protect the torso and head from impact. The present results suggest that if this is true for a fall onto a soft surface, then it is most certainly true for a fall onto a hard surface because there is even less time for the reflexes to have any meaningful effect.

## 5.6 Acknowledgments

The authors wish to thank Kurt M. DeGoede, Ph.D., for conducting pilot studies (DeGoede et al., 2002) leading to this paper. We thank the subjects for their participation and the financial support of PHS grant P30 AG 024824 is gratefully acknowledged.

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# CHAPTER 6

## General Discussion

### 6.1 Novel insights

This dissertation addresses a series of knowledge gaps pertaining to the behavior of the outstretched arm under the type of end-loading characteristic of protective bracing and/or arresting a fall to the ground. While the focus has been placed on forward falls because this is the most common fall direction (O'Neill et al., 1994; Nevitt and Cummings, 1993), many of the same issues hold for arm behavior during lateral or backward falls when an arm is used for protection. There is a conundrum of course. If the arm is purposely hyperextended so as to prevent any risk for elbow flexion buckling, then this 'outstretched arm' strategy increases the risk for a wrist fracture. But if the arm is held in slight flexion by arm muscles that are not preactivated enough, or the arm muscles are too weak, or the arm is too flexed, then the arm will buckle, increasing the risk for a head or torso injury.

These are the first experiments to use impulsive elbow flexion moments, applied via a transverse force to the distal forearm, to identify the elastic and viscous resistance of the actively contracting elbow muscles of young male and female volunteers to small sudden stretches (Chapter 2). These order of magnitude estimates for the elbow extensor muscles then served as a useful starting point for examining the corresponding values found in a more innovative and more realistic experiment in which subjects had to adopt the same posture as in a forward fall arrest while resisting impulsive end-loading to the wrist (Chapter 4). In this experiment optimization was then used to identify the elastic and viscous coefficients of the elbow and the shoulder muscles when the upper extremity of young and older volunteers was end-loaded by modest impulsive compression forces at the wrist. The innovation was in obtaining these estimates in a safe manner without incurring any adverse events such as soft or hard tissue injuries whether young or old. We found that age, gender, and the pre-contraction level affected the viscoelastic properties of the precontracted elbow extensor muscles, but we did not find significant gender or age effects on the shoulder muscles. However the results of the shoulder studies imply that the initial elastic properties of the shoulder extensor muscles play an important role in controlling the responses of upper limb to impulsive end-loading (Chapter 4).

The most useful insights from this dissertation include the first estimates of how age, gender and pre-contraction level of the arm and shoulder extensor muscles affect the behavior of upper extremity when resisting an end-load such as when bracing against the dashboard in a frontal vehicle collision or when arresting a fall. The simulation study (Chapter 3) is the first to identify the buckling load for an end-loaded arm, and the first to show that the predicted critical buckling load for older females in a forward fall was 40% of that of young males having a similar body size. The most important insight is relating the buckling load to arm muscle protraction strength because this has practical implications for training the elderly. We found that the elbow muscle elasticity and damping were proportional to muscle strength; however we did not find a strong relationship between shoulder muscle stiffnesses and muscle strength in the experimental studies (Chapter 2 and 4). The former result helps explain why older women are susceptible to head injuries during falls and underlines the importance of maintaining upper extremity muscle strength throughout the age span.

We were able to explore the estimates of the buckling load and the effect of age and gender on this buckling load under sudden stretch of a human extremity muscles without risk of injury by using computer simulation models (Model I in Chapter 3). We added the non-linear rotational stiffness and damping values to explore how the torque-angle curve shape of the elbow and shoulder extensor muscles could affect the behavior of upper limbs under sudden increase in end-load (Model II, Chapter 3). Furthermore we then expanded the 2-D model to a 3-D model of the arm and shoulder to predict how shoulder flexion/extensor muscle and ad-/abduction muscle properties affect the arm load-deflexion behavior to changes in the peak load on the hand (Model III, Chapter 3).

The simulation Model I in Chapter 3 provides the first estimates of the critical load and the critical elbow angle based on anthropometric data such as body weight and height and/or arm muscle maximum voluntary strength (MVS) using a simple planar model including elbow and shoulder. Model II in Chapter 3 predicted the non-linearity of the torque-angle measurement shape on the elbow and shoulder extensor muscles affected adversely the critical load therefore the critical load with the non-linear condition reduced 16% less than with linear condition. A schematic torque-angle relationship for the elbow extensor muscle is shown in Figure 6.1 (Carpes et al., 2012, Piitulainen et al., 2013). The middle curve ("1.0*MVS") is normally measured as the maximal voluntary isometric strength in the laboratory. The upper curve ("2.0*MVS"), essentially double the middle curve's values, can be estimated from knowing that a fully activated muscle that is stretched dynamically can maximally develop twice the maximum isometric contraction force (for example, Edman, 1988; Rassier et al., 1999). The lower curve ("0.5*MVS") is a safe starting load to start a standard progressive resistance training program wherein one lifts a weight that is half one's maximum strength in three sets of 10 repetitions (White et al., 2004).

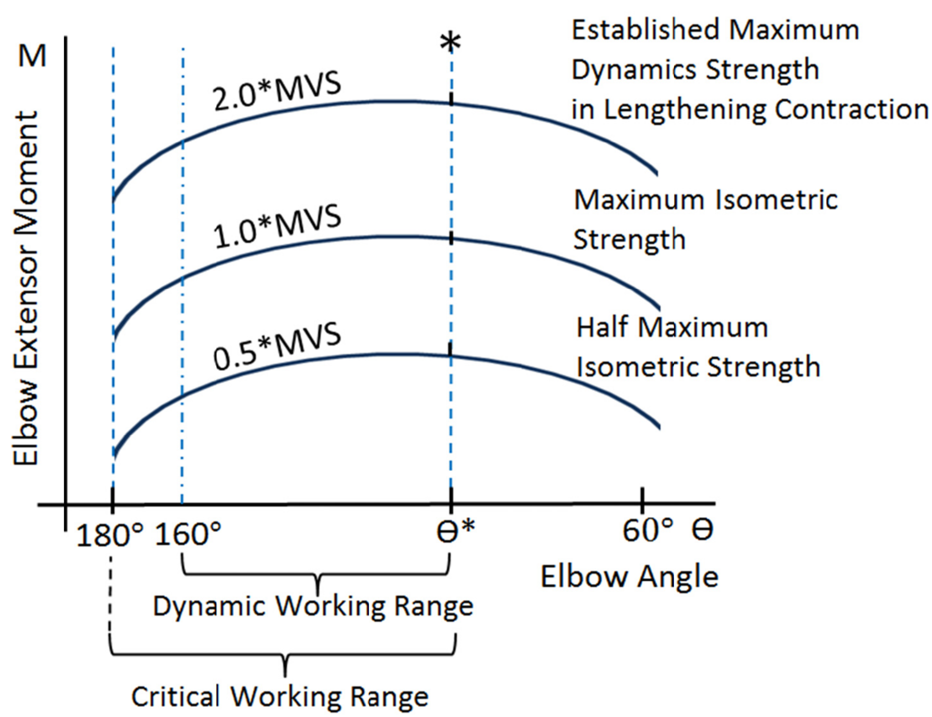


**Figure 6.1** A simplified torque-angle relation for elbow extensor muscles. An elbow angle of 180° is fully extended.

If a grandmother weighs 60 kgf, then we can estimate her critical arm end-load as 350 N, based on the results from the simulation study (Chapter 3). This load would establish the requisite dynamic elbow strength needed at an elbow flexion angle of 65 degrees when landing a fall (DeGoede and Ashton-Miller, 2002) in Figure 6.1; then half of this load, namely 175 N, could be used as the weight to eventually be resisted repetitively in progressive resistance training of her elbow extensor muscles (for example, Claflin et al., 2011).

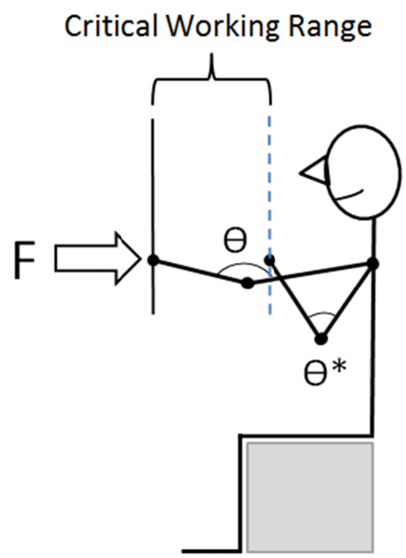


**Figure 6.2** An example for training elbow extensor muscles in the sitting position. Catching a medicine ball thrown at one's chest would be one example of such a task.

In Chapter 1 Figure 1.8, we conceptualized the reaction to arresting a fall when the hand strikes the ground and the elbow angle flexes from the initial angle (Θ) to the critical angle (Θ*). This usually happens in a short time span, under 150 msec (for example, Dietz 1981; DeGoede et al., 2003; Sran et al., 2010; Burkhart and Andrew, 2013; Chapter 5). However we can more safely train the arm and shoulder muscles under more controlled circumstances with a fraction of the predicted critical load.

Figure 6.2 shows one example for training the arm extensor muscles under an impulsive load on the hand when in the sitting position. If one knows the critical load for an older adult, then we can train him or her effectively with half the critical load between outstretched elbow angle and the critical elbow angle to establish dynamic strength for her upper extremities (Figure 6.2). Since older muscle is injured more easily in a lengthening (eccentric) contraction (for example, Brooks et al., 1995), one needs to progressively train the arm extensor and protraction muscles under lengthening contraction conditions, hence the loading involved in catching a medicine ball thrown towards one's chest (Figure 6.2) would be a useful training stimulus. An alternative training scenario would be to have the subject decelerate the bob of a pendulum swinging towards the chest, as used by DeGoede et al. (2002). This would also constitute a favorable stimulus for the bones of the upper extremities because the cells of both cortical and cancellous bone responds and remodel well to high frequency compressive stress (for example, Rubin et al., 2002).

The propagation of an impulse along the upper extremity is such that there is not sufficient time for muscle reflexes to significantly increase muscle resistance to arm buckling during a fall (Chapter 5). This underlines the importance of setting an adequate level of muscle pre-contraction in the upper extremity prior to hand contact with the ground or the onrushing object whenever the arm is used to protect against head or torso injuries.

An ultimate goal of falls biomechanics research can be to develop comprehensive training guidelines for teaching the older adult for how to arrest a fall safely and where possible training them to ensure that they have adequate physical capacities to achieve this. The insights obtained from this dissertation should aid this endeavor. Future research is warranted to design, develop and test a training program. But based on the present findings, we believe that learning to avoid hyperextended arms when possible, learning to set an adequate pre-contraction level in the arm muscles to ensure sufficient resistance to buckling, learning which exercises will help strengthen the arm protraction muscles, and doing those exercises regularly are keys to safely arresting falls.

### 6.2 Straight vs. slightly flexed upper extremity loading configuration

Why does landing on a hard surface with a straight arm increase the risk for wrist fracture? There are two main reasons. First, a negative effect is structural stiffness of the arm in the direction of end-loading essentially becoming that of the lower and upper arm bones under compression. If bone experiences 3,500 microstrain (0.35 %) (for example, Burkhart et al., 2012) under a compressive impulsive load of 1*BW, say, and if BW is 750 N, then a 50 cm-long upper extremity, stiffened by a collinear radius (with ulnar) and humerus, will deflect 1.75 mm; that number would represent a geometric stiffness of 429 N/mm in compression. If, on the other hand, an elbow flexes 10 degrees under impact, then the change in effective length of an arm whose segments each measure 25 cm would be (25.0x(1-cos ($10^{o}$)) = 3.8 mm or twice as much deflection at the wrist; if the elbow flexed $20^{o}$ under the impulsive load, the change in overall length would be 1.5 cm, equivalent to nearly an order magnitude lower geometric stiffness than that of the boney structure. The more the elbow flexes, the more time there is for the downward momentum of the arm segments to be arrested, the lower the peak impact force is and the more energy that can be absorbed by the negative work done by stretching the extensor muscles acting about the elbow.

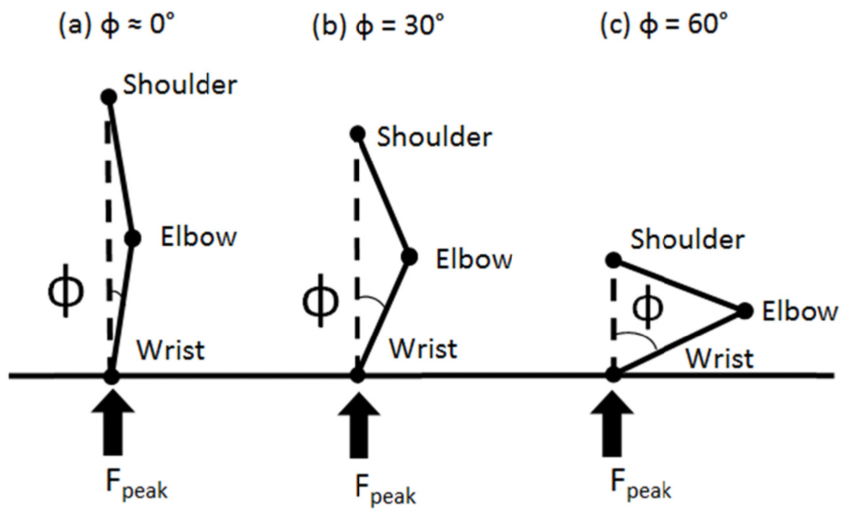


**Figure 6.3** An estimation of skeletal and muscular components of upper limb stiffness calculated as a function of the angle between $F_{peak}$ and the arm with : $k = k \cos^2\phi + k \sin^2\phi$, where, k = observed upper extremity stiffness, $\phi$ = the angle between $F_{peak}$ and the arm, $k \cos^2\phi$ = skeletal component, and $k \sin^2\phi$ = muscular component (after DeVita and Hortobayi, 2000).

The estimation of the upper extremity stiffness in the present study should be considered as the muscular component rather than an assessment of the skeletal contribution to resist against collapse (DeVita and Hortobayi, 2000). This approach could explain why women had the lower deflexion angles in elbow and shoulder joints than in men (Chapter 1) though we found that women have the lower stiffness and damping coefficients of upper extremities. Using DeVita and Hortobayi (2000) estimation of the skeletal and muscular components of lower limb stiffness, we could represent the upper extremity stiffness as a function of the angle between $F_{peak}$ and the arm with $k = k \cos^2\phi + k \sin^2\phi$ where, k = observed upper extremity stiffness, $\phi$ = the angle between $F_{peak}$ and the arm, $k \cos^2\phi$ = skeletal component, and $k \sin^2\phi$ = muscular component (Figure 6.3). A completely extended arm would have $\phi = 0$ and all stiffness depends on the stiffness, k, of the skeletal component in Figure 6.3 (a). As elbow flexed, the muscular component increases ($\phi > 0$) in Figure 6.3 (b) and (c), therefore the relative contribution of skeletal to muscular components is decreased.

With the above geometric stiffness 429 N/mm, the structural stiffness abruptly decreases as the elbow flexion angle increases (black diamonds in Figure 6.4). Then, we estimate the muscular stiffness in active responses from the rotational stiffness data at elbow from Chapter 4 to the muscular linear stiffness which is about 2.4 N/mm in average as shown in blue circles in Figure 6.4. We can see the much smaller values in the muscular stiffness rather than the structural stiffness under one body-weight critical load condition. This relationship could explain why the women with advancing age increase their dependence on the skeletal system and reduce their dependence on the muscular system under an impulsive end-load and this is the reason they need to strengthen their muscular capacity.

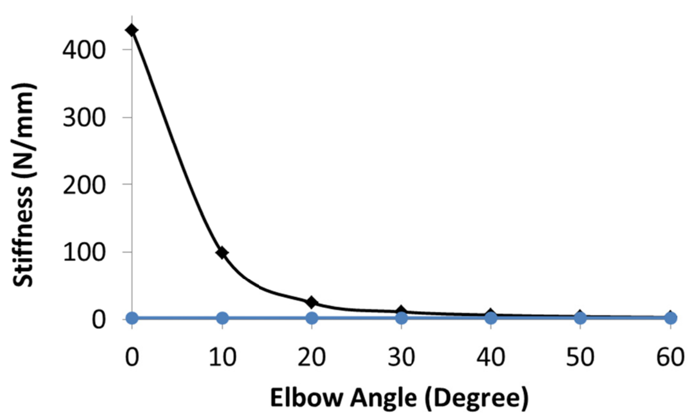


**Figure 6.4** An estimation of structural stiffness (in black diamonds) and muscular stiffness (in blue circles). This is an example of simple calculation with the buckling load = 750 N, the arm length = 25 cm, and the muscular stiffness is calculated from the rotational stiffness results at elbow from Chapter 4.

The second negative effect of hyperextending the elbow in preparation for wrist impact with the ground is that instead of just the segmental mass of the forearm being the mass of the presenting body segment to the ground, the hyperextended elbow tightly couples the segmental masses of the upper and lower arm together so as to double the mass of the presenting body part to that equal to the forearm and upper arm. This then doubles the momentum of the presenting body segment that has to be arrested by the ground reaction force over 10-15 ms. The result, when landing on a hard surface, is much larger impulsive force sufficient to fracture the wrist (for example, Frykman, 1967). So hyperextending the arm in preparation for landing a fall is a risky strategy for the wrist, as well as the rotator cuff of the shoulder (for example, Lungren et al., 2006) although if one has to choose between a broken wrist and a head injury, the former may the least of the two evils and this may be a strategy adopted by some as a last resort. This dissertation did not examine falls with hyperextended arms, but instead focused on the question of how much strength and muscle activation is needed to prevent the slightly flexed arm from buckling in men and women of different ages.

### 6.3 Relationship between Arm Buckling Load (F*) and Arm Extensor Strength

This dissertation provides useful insights for designing exercise and rehabilitation goals for the upper extremity strength and pre-contraction level required by older adults, especially women to land a fall safely (Chapter 6.1). Based on experimental (Chapter 2 and 4) and simulation (Chapter 3) results, we predict a proportional relationship between arm buckling load (F*) and arm extensor strength.

How many push-ups one can do is one simple assessment of the upper body muscle strength (for examples, Contreras et al., 2012; Suprack et al., 2011; Cogley et al., 2005; Wood et al., 2004; Baumgartner et al., 2002; Chou et al., 2002; Vossen et al., 2000; Mayhew et al. 1991; Dean et al., 1987). Dean et al., (1987) found that the push-up times body weight was a good predictor of muscle strength and Vossen et al., (2000) choose the medicine ball throwing for training of upper extremity strength. Therefore if we know the number of push-ups, we can predict the upper body strength as well as the best load for arm protraction strength training.

One biomechanical study of the standard push-up found that it required 66% MVC of triceps (for elbow extensor) and 61% MVC of pectoralis (for shoulder adductor) and 42% MVC of anterior deltoid (for shoulder flexor) in college-aged men and women (Contreras et al., 2012; Baumgartner et al., 2002). Similarly, Cogley et al., (2005) found pectoralis major activity of 64% MVC in young men and 106% MVC in young women; and triceps brachii activity of 69% MVC in young men and 113% MVC in young women. So the women needed to use significantly greater magnitudes of EMG activation than the men, and that activity exceeded their maximum strength. Therefore, the knee push-up is a better arm extensor training activity for women to start with because the "knee push-up" reduces the percentage of body weight that has to be supported by the hands to 54% in the fully extended arm ("top") position and 62% in the flexed arms ("bottom") position (from 69% of body weight in top position and 75% of body weight in the bottom position) during the standard push-up (Contreras et al., 2012; Suprak et al., 2011; Wood et al., 2004). It may then be wise for women to start out with the knee push-up to build arm extensor strength until they can do at least one standard push-up on hands and toes. One can estimate from the above results that a female requires 93% of MVC triceps effort to do a knee push-up since she supported 75% body weight in the "bottom" position during the standard push-up and the support was reduced 62% body weight during a knee push-up.

**Table 6.1** Comparisons of measured and estimated loads on hand during a 70 cm fall arrest, Model I predicted buckling load, and a single push-up.

| | Young Male | Young Female | Old Male | Old Female |
|---|---|---|---|---|
| (A) Measured body weight (BW) | 72 kgf (DeGoede 2002) | 60 kgf (Case 2005) | 69 kgf (Lee 2013) | 68 kgf (Lee 2013) |
| (B) Measured peak load on each hand during an 70 cm an fall arrest | 560 N (DeGoede 2002, 75 cm Fall) | 480 N (Case 2005, 70 cm Fall) | 537 N (= 560/720×690) (ratio estimated from DeGoede 2002) | 544 N (= 480/600×680) (ratio estimated from Case 2005) |
| (C) Load on each hand in "down" position of standard push-up (75% Body Weight, Suprak 2011) | 270 N (=0.75 x 720 / 2) | 225 N (=0.75 x 600 / 2) | 259 N (=0.75 x 690 / 2) | 255 N (=0.75 x 680 / 2) |
| (D) Maximum dynamic end-load if only a single push-up ispossible (estimated as 2x(C), See Figure 6.1) | 540 N | 450 N | 518 N | 510 N |
| (E) Model I predicted buckling load | 715 N | if BW = 60 kgf, 500 N | if BW = 69 kgf, 600 N | if BW = 68 kgf, 400 N |
| (F) Predicted arm buckling? (i.e., (B) > (E) ? ) | No | No | No | Yes |
| (G) Arm buckling observed in experiment? | No (DeGoede 2002) | No (Case 2005) | NA | NA |

How do the arm buckling loads predicted by Model I compare with the dynamic loads on the hand during a forward fall to the ground from 70 cm shoulder height, and the loads on a hand during a standard push-up? Table 6.1 makes this comparison for healthy young and older females, and healthy young and older males. The results tell us that the older male arm should not buckle a forward fall of the severity considered. But Model I tells us that for an older female of body weight 72 kgf, her arm would buckle when the impulsive end-load exceeds 400 N while the young female having the same body weight should buckle under a 600 N end-load. Case et al., (2005) measured 482 N peak GRF when young females fell forward from a shoulder height of 70 cm, 584 N from 80 cm, and 646 N from 90 cm fall height using the DeGoede fall protocol (DeGoede and Ashton-Miller, 2002). In Table 6.1 we see that Model I would not predict that either the younger female or male arm would buckle during a 70 cm fall arrest, and that is indeed what was found by DeGoede et al., (2002) and Case et al., (2005). The estimates in Table 6.1 suggest why the young women in the latter study severely limited the increase in elbow flexion during all fall arrests, but particularly the 90 cm: because the impulsive end-load was so close to the buckling load of their arm, as predicted by Model I. But the story is very different for the older women. In Table 6.1 we see that the predicted buckling load is significantly (~22%) less than the estimated dynamic end-load on the arm during a fall. We see in Table 6.1 that if one doubles the load on the hand during at the "down" position of a standard push-up, one can estimate the maximum dynamic load that the arm can tolerate if the protractor muscles are rapidly stretched (Figure 6.1). These "dynamic protraction strength" estimates are similar to the peak force on the hand during a 70 cm fall for the younger adults, and so confirm how much arm strength and stiffness are needed to prevent buckling. But only the few older women who can still do a single standard push-up, without having gained any weight over their lives, might prevent the dynamic ground reaction force causing arm buckling in a 70 forward fall (i.e., 510 vs 544 N, Table 6.1). However, because it predicted a buckling force in the older female arm of only 400 N, Model I is pessimistic in this regard. The results of Lo (2008) suggest that landing on one or both knees first is one way to reduce the dynamic end-load on the arm in a forward fall and, as unpleasant as this can be, it is one strategy to prevent the end-load from reaching the critical value that would buckle the older female arm, given limited protraction strength capacity.

Apart from the question of training arm *strength* to prevent arm buckling, might older adults have to train the *speed* of their reactions to an impending fall? First, the time it takes to fall depends on how much initial forward momentum one has at the start of the fall. From DeGoede (2000, Figure 5) we can estimate that a younger person walking at 1.24 m/s will strike the ground in 560 ms, whereas an older person walking at 1.11 m/s will strike the ground in 650 ms. This turns out to be plenty of time to deploy a protective response with the arms. For example, in a sport-protective task aimed at testing how quickly one can raise the hands against gravity to protect the head from an oncoming object, Lipps et al. (2013) found reaction, movement and total response times of 71, 143 and 14 ms, and 79, 163 and 242 ms in healthy young males and females, respectively. As to the effect of age, DeGoede et al. (2001) found protective hand-arm movement times of 226 and 285 ms in young and older males, and 274 and 297 in young and older females. So whether young or old, male or female, one has plenty of time to get the hands into a protective posture before ground impact. So training *response times* in older subjects would not appear to be necessary. In part this is probably because raising one's hands to protect the head or torso when threatened is an overlearned response because it is used so frequently over the life span.

## 6.4 Limitations of the Approach

Since the possibility of injuries, for example bone fracture, bruise, rupture of muscles or ligaments on upper extremities under an impulsive end-loading, the peak load condition *in vivo* studies (Chapter 2, 4 and 5) were modest compared to those in a real forward fall event. The peak impact force in the studies with healthy young adults from standing height (Dietz et al., 1981) or from the waist height (DeGoede and Ashton-Miller, 2002) reached more than 1,000 N on hands. The peak force in other experimental studies was 200 N on hands when subjects were dropped from 10 cm height (Burkhart and Andrew, 2013). Sran et al. (2010) studied the age difference in energy absorption of the descent phase movement from different forward body lean angles; the peak hand force was 170 N in young adults and 120 N for older adults. There is no study with the peak hand forces over one body weight (700 N) for the elderly. Therefore the relatively modest load condition (250 N) we used on the hands was safe, especially for our healthy older subjects. It sufficed to permit estimates of the rotational viscoelastic properties of pre-tensed arm and shoulder muscles and these were affected by age, gender and pre-contraction level of arm muscles.

Had we employed the impulsive compression forces of a magnitude that occurs during real falls, say four times larger than we used in the present dissertation, we would expect the values of elbow and shoulder elastic and viscous coefficients to be larger, but perhaps not that much larger. This is because the maximum amount that an individual can increase the stiffness of the muscles about a joint is about three-fold (for example, Blanpied and Smidt, 1993). But since the present subjects were already pretensing their arm extensor muscle to 75% of their maximum values (Chapter 2 and 4), we do not expect the coefficients would be more than 1.25 times larger. So our present estimates are underestimates of these values.

A second limitation is that the experiments of Chapter 4 were conducted without the subject actually falling. While the subjects did have to pretense in anticipation of the impact, we cannot be sure that they were as aroused as they might have been in a real fall scenario. So the postural set might have led the coefficients we estimated are underestimated.

A third limitation in the approach we adopted for the computer and experimental studies in this dissertation is that the falls were considered sagittally-symmetric. Since asymmetric responses may be common than symmetric falls (Lo, 2006), we decide to test with non-dominant arms *in vivo* studies for the worst-case scenario so our results might be underestimated for the dominant arm. However we felt that it was necessary to understand the kinematics and dynamics of the symmetric fall in the computer studies before trying to tackle the kinematics and dynamics of asymmetric falls.

A fourth limitation is that we only used lumped parameter models for the upper extremity (Chapter 2-4). There are other modeling approaches such as the OpenSim approach in which each muscle is represented (for example, Seth et al., 2011). The problem with those approaches is that there are no models of how a whole active muscle responds to a lengthening or eccentric contraction. So even if one faithfully represented each muscle and its line-of-action using the OpenSim model, the model would not necessarily be any more accurate in predicting the elastic and viscous resistance of those arm muscles to sudden stretch. A related limitation is the fact that we did not couple the resistance provided by the muscles acting about the elbow to those acting about the shoulder, whereas in reality there are two joint muscles like the long-head of biceps and triceps that cross both joints. In our modeling we considered the two sets of coefficients (those for the elbow and shoulder as independent).

A fifth limitation is that the mechanics of the neck, the torso, and the translational shoulder movements were not represented in the model simulations. For example, neck flexion will increase the critical shoulder height (h*) and the critical elbow angle (ϴ*) needed to ensure that the head does not strike the ground even in the presence of excessive neck flexion (Figure 6.5). The posterior translational shoulder motion in the transverse plane that affects the shoulder girdle muscles was also not considered in the present simplified simulation models.

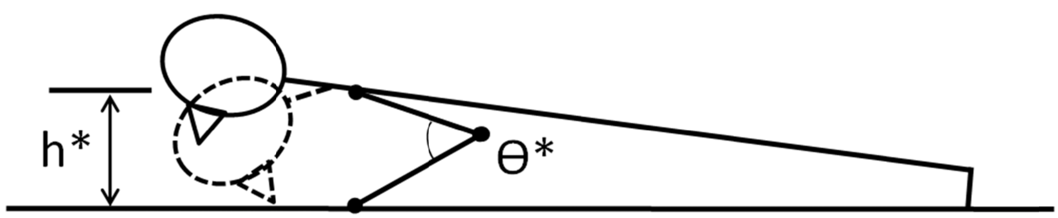


**Figure 6.5** Schematic diagram showing how excessive neck flexion (dashed line) can cause the head to strike the ground even if the downward motion of the shoulders is arrested in time during a fall to the grounds (solid line). The shoulder height, h and the elbow angle, ϴ would change to the critical shoulder height, h* and the critical elbow angle, ϴ* when arm buckled (or head hit) with the fully flexed neck posture. So, for no head impact to occur under this scenario, h* and ϴ* should ideally be set large enough to exclude the possibility of head impact even with excessive neck flexion. Because we did not study actual falls or the range of available neck flexion in this study, we cannot predict how much larger h* or ϴ* should be in a fall to the ground to avoid head injury under any scenario.

A sixth limitation is that the models used in Chapter 3 do not represent the soft tissue packages (for example, Terroso et al., 2013). These are used to represent the mass distribution of the muscle and fat and skin of the upper extremity and their compliant coupling to the bones of the upper extremity. In Chapters 2, 4 and 5 we used active markers attached to the soft tissue packages so, because of this compliant coupling, these kinematic measurements would systematically underestimate the initial transfer of momentum to the arm bones (Chapter 5). In Chapter 3 we assumed that the momentum of an entire arm segment had to be arrested in a certain time interval whereas in reality the momentum of the bones have to be arrested in that time interval and the momentum of the soft tissue packages (if they were present) would be arrested in a slightly longer window, because of compliance in their coupling to the bones. Therefore we may have overestimated the viscoelastic values (K and B) for the muscles about the elbow and shoulder joints in Chapter 2 and 4, the propagation time of impulse along the upper limbs in Chapter 5, and the main outcomes of buckling load and underestimated the joint deflexion angles in Chapter 3 by not including the effect of the soft tissue packages.

A seventh limitation is that no actual falls were studied in this dissertation. Perhaps it is innovative that the use of other experiments and computer simulations replaced the need to study actual falls with their risk for injury. But sooner or later one needs to validate the models to make sure that the predictions are correct. The best validation would be to use actual falls to do this. But the problem is how to do this responsibly without injuring the older adult either overtly or covertly. In the case of the latter one worries about causing occult rotator cuff microinjuries that only later coalesce to cause a rupture of an important structure comprising the rotator cuff. It is possible that

DeGoede's (2002) experiments could be replicated in older adults who are physically fit and who have been carefully screened to being doing recreational activities in which they fall now and then, such as skiing, skating or martial arts, for example.

An eighth limitation is that there were no independent estimates of muscle tissue elasticity or viscosity. These would have helped validate the methods. Ultrasound shear wave elastography (Eby et al, 2013; Greenleaf et al., 2003) offers promise in this regard, being a non-invasive technique for measuring muscle stiffness at various levels of activation. A method would have to be found to relate the elasticity in shear to that in tension.

A ninth limitation is the dearth of information on the force-length-time behavior of a whole striated muscle under rapid stretch. The computer simulations in Chapter 3 were based on data from Grover et al (2007), but it would have been sensible to run sensitivity studies to examine how sensitive the results are to changes in the breakpoint of the bilinear curve, as well as the slopes of the bilinear relationships. If the breakpoint occurs earlier than at 14% of the range of motion on the Grover curve, there was a large effect on the critical load, so the torso and head would be more vulnerable than before.

A tenth limitation was that only one of the two upper extremities was impulse loaded in Chapter 4. While it was interesting to measure how a single extremity responded to impulsive loading, the resulting motion of the head, neck and torso was different than in a bimanual arrest because the latter would involve less axial torsion of the body due to the asymmetric impulse loading.

An eleventh limitation is that it is important to consider the gender difference in body mass distribution which can occur with advancing age. For example, men tend to put weight on their stomachs, while women tend to put weight on their thighs as they grow older. So, men having the same height and weight as women would tend to place a larger dynamic load on their hands in a fall because of their higher center of mass (for example, Smith et al, 2002). However, our present estimates on buckling load in older females have taken the gender difference in arm mass distribution into account.

In summary, despite the known limitations of the *in vivo* experiments and the limitations of the upper extremity model simulations, the main findings of this dissertation should be of the right order of magnitude. The methods were reliable enough to estimate the viscoelastic values for the arm and shoulder active responses, the critical (buckling) load on hands under an impulse, and how muscle strength affects this value. These findings provide a framework for better understanding of how biomechanical factors determine whether or not an arm will buckle when end-loaded in a fall arrest.

### 6.5 Recommendations for Future Research

Many unresolved questions remain regarding the biomechanics of the arrest of falls:

(1) The experimental studies in this dissertation involved studying the response of a single upper extremity to impulsive end loading. The computer simulations also studied the predictors of single extremity buckling. But the focus was on the postures adopted in bimanual forward fall arrests such as those studied by DeGoede et al. (2002). Clearly more work needs to be done to study how the single end-loaded extremity and torso responds in lateral or backward or even single arm fall arrests

because this affects the direction of loading of the shoulder and which shoulder muscles will be lengthened. Since the shoulder has different strengths in flexion, abduction and extension, this will also affect its rotational and translation elastic and viscous resistance to the impulsive loading.

(2) Additional experimental studies of actual forward fall arrests are needed, if they can be done safely, to validate the model predictions.

(3) Simulation studies including the dynamics of the neck, the torso, and the translational shoulder movements need to be conducted. As mentioned that the neck flexion could be crucial at the critical elbow angle (see Figure 6.5), the biomechanics of the neck including viscoelastic properties need to be studied. The posterior translational shoulder motion is controlled by the shoulder girdle muscles and these should be represented in future models.

(4) Additional theoretical and experimental studies need to be conducted to explore the human upper body response in other fall directions, such as lateral, posterior, posterolateral, and anterolateral directions. Are older adults capable of configuring their bodies during a lateral fall so as to avoid a hip impact, perhaps by using both arms? Is there a universal and safe fall arrest strategy which can be applied to all fall directions and can be performed by young and older adults with ease?

(5) More detailed modeling of whole muscle response to rapid sudden stretches is needed, along with experimental validation. These should be conducted with muscles having different architectures and preset activity levels. The behavior of the muscle-tendon unit needs to be quantified.

(6) Finally, an intervention program aimed at teaching people how to protect themselves from fall related injury might be developed based on insights from these biomechanical simulations and experiments (for example, Figure 6.2).

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# Chapter 7

## Conclusions

(1) Healthy young women exhibited 72-100% of the elastic resistance and 42-76% of viscous resistance when their pre-activated elbow extensor muscles were suddenly stretched under a standardized step increase in elbow flexion moment (Chapter 2). The results of the lower maximal elbow extensor stiffness in women than in men of the same size suggest that women has smaller capacity to prevent elbow flexion buckling under fall-related impact than men therefore, women need to maintain elbow extensor strength with age (experimental study in Chapter 2).

(2) A new developed apparatus was used to measure the effect of age and gender on the elbow extensor and shoulder protractor muscle stiffness and damping values under an impulsive end-load at the wrist (Chapter 4). Healthy older adults exhibited 57% − 95% and 76% of healthy young adult normalized stiffness and damping values; healthy women had 51% – 80% and 66% of healthy men's normalized stiffness and damping values at the elbow.

(3) Pre-contraction level significantly affected the rotational stiffness and damping resistances of the upper extremity joints: the elbow (Chapter 2) and the shoulder in sagittal and transverse planes (Chapter 4) when the arm was suddenly end-loaded at the wrist. Stiffness values significantly increased with the muscle pre-contraction level, as well as with the elbow initial angle. Damping values varied with pre-cocontraction level, not by age nor gender.

(4) A biomechanical computer model was used to predict the buckling load and the critical elbow angle (Chapter 3, Model I). The buckling load of healthy young man was equal to the body weight, while the buckling load of older women having the same body weight and height was predicted to be 60% of the young male value. The effect of initial elbow angle helps explain why an older female may strike their head more easily than a young male in arresting a forward fall.

(5) The arm buckling load was found to be more sensitive to the elbow viscoelastic properties than the shoulder's properties (Chapter 3, Model II). The buckling behavior was sensitive to the initial elbow angle at impact; the greater initial elbow angle, the greater elbow deflexion angle. The larger non-linearity relationship of the torque-angle shape on the muscle stretch predicted the less capacity of the arm buckling load which is unsafe.

(6) Shoulder adductor muscle properties significantly affected the shoulder deflexion angle as well as the elbow deflexion angle, but neither the shoulder extensor

muscle properties nor the shoulder adduction angle affected the elbow deflexion angle (Chapter 3, Model III). This study suggests that it is wise to maintain good shoulder adductor strength as well as triceps strengths in order to avoid limb collapse and the head striking the ground in a forward fall.

(7) The experimental measurements at the shoulder joint emphasized the importance of the elastic resistance in the flexion and extension muscles at the shoulder. From the experimental and theoretical studies on the two different shoulder planes (Chapter 3 and 4), we conclude that the viscoelastic properties of the shoulder muscles in both sagittal and transverse planes are important to arrest a forward fall.

(8) The bilinear characteristic in muscle tensile stiffness was utilized in Chapter 4 to estimate stiffness values at the elbow and the shoulder joints, based on Chapter 3 Model II results. The bilinear stiffness values were found suitable for the Chapter 4 computer simulation to optimize the values of the two stiffnesses $K_1$ and $K_2$ at the elbow and shoulder across all subjects.

(9) The propagation of impulse along the upper extremity was sufficiently rapid that no neuromuscular reflex can augment arm or shoulder muscle stiffness in time to prevent arm buckling under an end-load (Chapter 5).

(10) In terms of clinical impact of this dissertation, the positive relationship between the buckling load and the muscle strength of upper body found in Chapter 3 raises the possibility of using the push-up as a test in the clinic, or a self-test at home. For example, the results suggest that in the young adult, being able to execute a single standard push-up ensures that a subject has sufficient arm strength to prevent arm buckling in a fall, whether male or female.

(11) In Chapter 6 (General Discussion) it is argued on the basis of the findings in this dissertation, and biomechanical studies of push-ups in the literature, that males and females should endeavor to maintain the ability to do one or more push-ups for as long as possible as they age in order to maintain arm protraction strength. Even so, the results suggest that it will be challenging for older adults to prevent arm buckling under all but the most benign falls. However Lo (2008) showed that landing first on the knee(s) rather than the hands can significantly reduce the dynamic end-load on the arm in a fall, thereby reducing the likelihood of the arm buckling. This may therefore be a sensible fall arrest strategy for older adults to consider.

## 7.1 References

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