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The Relative Contribution of Human Cadaver Forearm Tissues to the Attenuation of Simulated Upper Extremity Impacts

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By

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ABSTRACT

This study aimed to determine the relative contribution of the forearm soft tissues (skin and adipose, muscle, interosseous membrane) to shock attenuation in human cadaver specimens following simulated forward falls. Peak acceleration decreased significantly as the shock wave travelled from distal radius (42.8 (19.0) g) to proximal ulna (12.3 (4.3) g) in intact human forearm specimens (n=9). Shock attenuation through the forearm decreased non-significantly compared to fully intact specimens when the soft tissues were sequentially removed (skin and adipose: 8.9%; muscle: 7.5%) and when the interosseous membrane was cut (21.9%). These results could help advance wobbling mass biomechanical models to study the upper extremity impact response of humans following forward falls. In doing so, our understanding of the injury mechanisms associated with impact-induced upper extremity injuries may be improved and strategies to reduce the number and severity of forward fall injuries may be realized.

DEDICATION

I dedicate my thesis to the 10 years of my university career and all the people who helped me achieve my goals. It has been a journey and I want to thank you for all the support.

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GLOSSARY

AC (acceleration slope): slope of the acceleration curve between 30% and 70% of the peak acceleration (measured in g/s).

Cadaver specimen: portion of a deceased human used in research.

Connective tissue: is a type of tissue that supports, connects or separates different types of tissues and organs of the body. It is made up of ground substances, fibres and cells.

CT: Computerized Tomography. Type of imaging system that produces cross-sectional images of different tissues of the body. Used for diagnostic and body composition analysis.

Effective mass: that portion of a body's mass that is involved in an impact.

In-vivo: experimentation that involves complete, living organisms.

In-vitro: experimentation that involves partial, non-living specimens.

Model: simplified version of a body, either in physical or mathematical form, that allows for simulations of the body in its entirety or parts.

PA (peak acceleration): peak acceleration magnitude as measured at the impact site following impact (measured in m/s^2). PA is typically expressed as a multiple of acceleration due to gravity, or g units.

Rigid segment: a body segment typically used in biomechanical modeling research which does not incorporate the motion of the soft tissues relative to bone.

SA (shock attenuation): a reduction in shock wave amplitude as it moves proximally through the body's tissues.

Shock: the transient condition that occurs following a sudden change in force application, causing the disruption of a system's equilibrium.

Shock wave: a stress or pressure wave that propagates through a medium.

Soft tissue: soft tissue includes tendons, ligaments, muscle, fascia, adipose tissue, membranes, blood vessels, nerves and skin.

Stress relaxation: under constant deformation, a material develops a high initial stress, which decreases as the deformation is maintained.

TPA (time to peak impact acceleration): time between impact and the peak acceleration (measured in ms or s).

Wobbling mass: includes all of the tissues (fat, muscle, skin) and fluids of the body that are not bone.

CHAPTER 1

INTRODUCTION

All age groups experience falls, but it is a more significant issue in the elderly due to unexpected disturbances in posture which occur as a result of tripping or slipping. Forward falls are the most common type of fall with more than half of the falls experienced by the elderly occurring in the forward direction (Nevitt & Cumming, 1993; O'Neill et al., 1994; Vellas et al., 1998). In forward falls, people often use their arms to brace for impact and protect their torsos and head (O'Neill et al., 1994; Hsiao and Robinovitch, 1998). Impact forces and subsequent shock waves are experienced at the hand and travel proximally through the forearms, elbows, arms, and shoulders (Burkhart & Andrews, 2010). These impact forces can cause serious injury such as distal radius fractures; 90% of these types of fractures are caused by forward falls (Chou et al., 2009).

Rigid (bone) and soft tissues (wobbling mass = fat, muscle, skin) have been shown to affect the body's response to impact including the soft tissue's ability to attenuate shock by 'wobbling', or moving relative to the underlying bone (Liu & Nigg, 2000; Pain & Challis, 2004; Chu et al., 1986; Hamill et al., 1995; Lafortune et al., 1996). As impact shock propagates through the body segments following impact, the rigid and soft tissues act to attenuate the amplitude of the waveform (Dufek et al., 2009; Voloshin & Wosk, 1982; Whittle, 1999; Wosk & Voloshin, 1981). Attenuation can be active, through changes in body kinematics and muscle activity (Cholewicki & McGill, 1995), or passive, through the movement of soft tissues relative to the underlying rigid tissue or bone (Chu et al., 1986; Hamill et al., 1995; Lafortune et al., 1996). Soft tissue oscillations that result from impact have been shown to be a significant contributor to the dampening of impact-induced shock waves (positive shock attenuation; reduction in shock wave amplitude as it moves proximally through the body's tissues) (Wakeling & Nigg, 2001).

Wobbling mass biomechanical models which incorporate both rigid and soft tissues have been shown to reduce impact forces compared to models comprised of only rigid segments (Gittoes et al., 2006; Gruber et al., 1998; Lui & Nigg, 2000; Pain & Challis, 2001, 2004, 2006). Schinkel-Ivy, Burkhart & Andrews (2012) quantified the effect that tissue composition of the leg had on the tibial response during impacts similar to those experienced during running. The authors found that the acceleration response of the leg (peak acceleration (PA), time to peak acceleration (TPA), acceleration slope (AS)) differed significantly depending on the ratio of rigid and soft tissue masses of the leg. Specifically, PA and AS decreased as leg lean mass and leg wobbling mass ratio increased. TPA increased as leg lean mass and leg wobbling mass ratio increased. Overall, the results of this study suggest that as soft tissue and rigid tissue mass ratio increase the acceleration response at the tibia decreases, indicating increased shock attenuation. Therefore, the different tissue layers have an effect on shock attenuation, but the relative effect of each tissue on shock attenuation remains unknown.

There have been different in-vivo and in-vitro methods used to research fall-related impacts. Burkhart & Andrews (2010) used a novel seated human pendulum device, designed to simulate the flight phase of a forward fall, to deliver bilateral impacts to the upper extremities of living participants. Chou et al. (2001, 2009) simulated forward falls to quantify the energy absorption abilities of the elbow and shoulder joints at impact, comparing the effects of relaxed and flexed muscles. Burkhart & Andrews (2011, 2013) used a Propelled Upper Limb fall ARrest Impact System (PULARIS) in a similar study to research the effects of different fall types and fall heights on the kinematics, kinetics and muscle activation levels of the upper extremity during a simulated fall.

With respect to in-vitro methods, Quenneville et al. (2010) developed and utilized a projectile system to test controlled impulsive loads over a large range of magnitudes and velocities to cadaveric, synthetic, or anthropomorphic test device tibias. The projectile system was modified in further research by Burkhart et al. (2011, 2013) in order to allow for cadaveric radius specimens to be tested in forward fall simulations.

Although a fairly extensive amount of research has already been conducted to analyze the effects of the different tissues of the lower extremity on shock attenuation (Schinkel-Ivy et al, 2012; Liu & Nigg, 2000; Wakeling & Nigg, 2001; Flynn et al., 2004; Holmes & Andrews, 2006; and Duquette & Andrews, 2010), relatively little has been conducted for the upper extremity in comparison. While numerous studies have shown the active force attenuating capabilities of muscle tissue through the use of models and in-vivo and in-vitro testing, the results from the Schinkel-Ivy et al. (2012) study clearly illustrate that individual soft tissue masses, especially lean mass, are important for shock attenuation. These findings support the need to quantify the effect that individual soft tissues have on shock attenuation for both the lower and upper extremities following impact. This thesis will begin this important investigation by considering the soft tissues (skin and adipose, muscle, interosseous membrane) of the forearm and their individual contributions to shock attenuation during impacts from simulated forward falls.

1.1 Research Question

This thesis will address the following research question:

What is the relative contribution of the skin and adipose tissues, muscle tissue, and interosseous membrane of the forearm on shock attenuation following simulated forward falls? This will be determined by a series of impacts to cadaver forearms following the removal of the different soft tissue layers, from superficial to deep.

1.2 Hypotheses

It is hypothesized that:

1. the shock wave (acceleration waveform) will be attenuated as it moves from the distal end of the forearm to the proximal end of the forearm. This attenuation will be characterized by a decrease in the magnitude of the peak acceleration and a positive shock attenuation value in the direction parallel to the long axis of the radius and ulna.
2. the attenuation between the distal end of the radius and the proximal end of the ulna will be greater than the attenuation between the distal and proximal ends of the radius when the interosseous membrane is intact. This is anticipated due to the damping effect of the interosseous membrane that occurs between the radius and ulna during forearm loading (Pfaeffle et al., 2000; Birkbeck et al., 1997). Harrison et al. (2005) showed that strain in the interosseous membrane increased with increased load at the wrist. When the interosseous membrane is cut, the load on the radius will not transfer to the ulna from the IOM and the attenuation between the distal end of the radius and proximal end of the ulna will increase even more.

3. as the layers of soft tissue are removed from the forearm, shock attenuation will decrease. Given the relative magnitude of the muscle mass, compared to the other soft tissues, the muscle tissue will have the greatest relative effect on shock attenuation. Schinkel-Ivy et al. (2012) found that local tissue composition of the distal lower extremity affected the shock attenuation at the proximal tibia. This study showed that as fat mass, wobbling mass, and bone mineral content increased, the acceleration response at the tibia decreased, thereby increasing shock attenuation. Therefore, as tissue layers are removed in the current study, shock attenuation should decrease, except when the IOM is cut.

CHAPTER 2

LITERATURE REVIEW

2.1 Tissue Properties

The characteristics of tissues affect how they respond to external loads, such as those experienced during a forward fall. The risk of injury from a forward fall is partially dictated by the characteristics of the tissues which comprise the body's segments. The rigid and soft tissues of the body can be characterized by the composition of the tissue or its response to loading. Stress (σ) is defined as the force per unit area in a structure, and strain (ε) as the deformation of a solid expressed relative to its original length. Human tissues exhibit a non-linear strain response to an applied stress, due to the high water content of most human tissue (Lu & Mow, 2008).

Given the inherent differences in material properties between tissue types, it is assumed that each tissue type plays a different role in shock attenuation (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010). Heterogeneity, anisotropy, and viscoelasticity are material properties exhibited by most human tissues (Whiting & Zernicke, 2008). Heterogeneity refers to the non-uniform composition of tissues, such as in cartilage, which is comprised of collagen fibres, and ground substance, which is made up of proteoglycans and elastin fibres. Anisotropy is the response of a tissue to a load which varies depending on the direction in which the tissue is loaded. Viscoelasticity refers to materials that elicit both elastic and viscous properties. Because of the viscous properties of tissues, their stress response is dependent on the amount and rate of strain imposed on the tissue. Viscoelastic tissues (soft tissues) exhibit behaviors such as creep, stress relaxation, and hysteresis. Tissues creep when they deform rapidly under an initial constant load, then continue to deform slowly while the constant load is maintained. Stress relaxation occurs when tissues undergo constant deformation and develop a high initial stress which decreases over time as the deformation is maintained Hysteresis is the energy lost as heat during deformation of a tissue, resulting in a delayed unloading pattern (Whiting & Zernicke, 2008).

2.2 Anatomy

The upper extremity consists of the hand, forearm and arm segments connected at the wrist joint, and elbow joints, respectively. Segments consist of soft tissues (skin, adipose tissue, muscle, and interosseous membrane (forearm)) and rigid tissue (bone).

2.2.1 Skeleton

Each upper extremity consists of 32 bones in the arm, forearm, and hand (Figure 1) (Tortora & Nielsen, 2014). The humerus makes up the brachium (arm) and is the longest bone in the upper extremity. It consists of the proximal head, which articulates with the glenoid cavity of the scapula, and the distal humeral condyles, which articulate with the radius and ulna of the forearm. The antebrachium (forearm) consists of the radius and ulna, with the radius being the major load bearing bone. The radius and ulna articulate distally with the manus (hand) via the scaphoid and lunate bones of the proximal carpus. There are eight carpal bones (scaphoid, lunate, triquetral, pisiform, trapezium, trapezoid, capitate, and hamate) arranged in two transverse rows of four bones each. The carpals articulate distally with the metacarpals, which in turn articulate distally with the phalanges of the digits. The metacarpals consist of five bones which have a proximal base, body and distal head. The fourteen phalanges of the fingers consist of a proximal row, middle row and distal row. The digits are numbered I to V starting from the thumb and moving medially to the little finger.

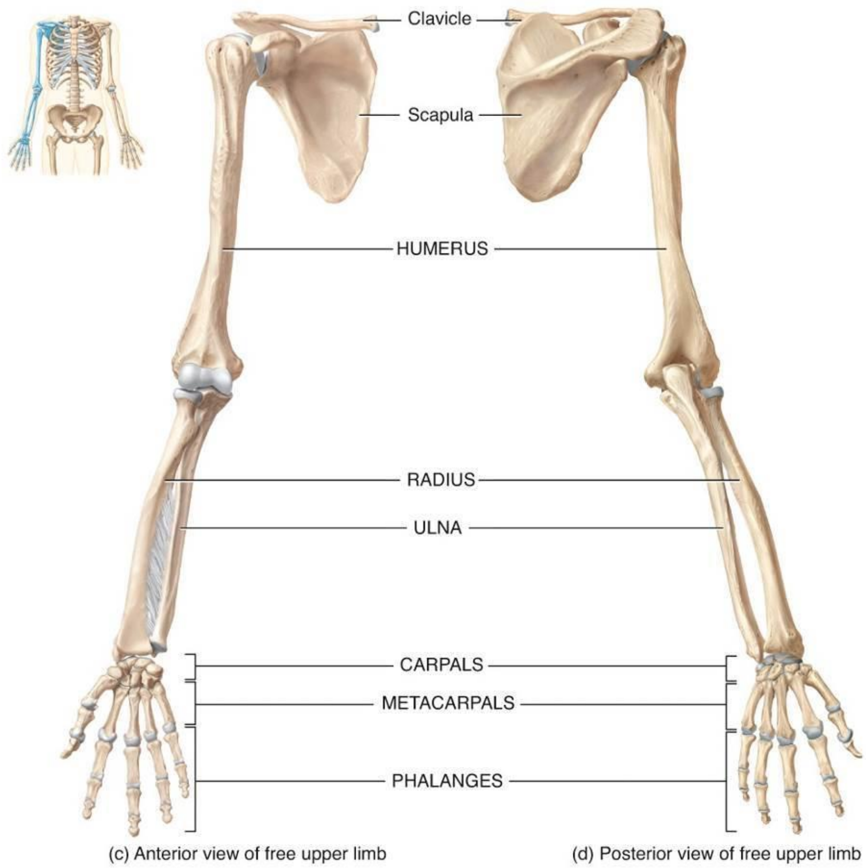

Figure 1: Bones of the upper extremity (Tortora & Nielsen, 2014).

2.2.2 Joints

The segments of the upper extremity articulate at predominantly synovial joints, which are characterized by fibro-elastic joint capsules and lubricating synovial membranes (Van De Graaff, 2000). The bones that articulate at each synovial joint are covered with a smooth layer of hyaline cartilage called articular cartilage. The ligaments that attach to the articulating bones provide a stable, flexible connection which helps bind the synovial joints together. The shoulder joint consists of the head of the humerus and glenoid cavity of the scapula. This ball and socket joint is multi-axial and provides considerable range of motion in all three major planes. The elbow joint, which involves the articulation between the trochlea of the humerus and the trochlear notch of the ulna, is classified as a hinge joint, permitting only flexion and extension. The proximal articulation between the radius and ulna is a pivot joint which allows supination and pronation of the forearm. The side-to-side articulation between the ulna and radius forms a syndesmotic joint held together by fibrous sheets called the interosseous membrane. The radiocarpal joint and metacarpophalangeal joints are condyloid joints, which are biaxial synovial joints that allow flexion, extension, adduction and abduction. The joints between the carpal bones of the wrist are gliding synovial joints which allow side-to-side and back-and-forth movements with slight rotation. The interphalangeal joints are hinge joints, allowing similar movements to those at the elbow joint. The joint between the trapezium and the metacarpal of the thumb is a saddle synovial joint which allows a number of movements including flexion, extension, adduction, abduction, circumduction and opposition (Van De Graaff, 2000).

2.2.3 Bone

Long bones such as the radius and ulna of the forearm, are composed of cortical and trabecular bone tissue. The bone mass of the human body consists of approximately 80% cortical and 20% trabecular bone, but trabecular bone constitutes 80% of the total surface area of bone (Teoh & Chiu, 2008). Cortical bone is comprised of osteons that resist bending and fracture via compression (Van De Graaff, 2000). Trabecular bone is a porous and spongy structure which constitutes the majority of the bone tissue internal to exterior layers of cortical bone, within short, flat and irregular bones and the epiphyses of some long bones (Van De Graaff, 2000). Trabecular bone forms a lattice network of rods and plates, which allow loads to be distributed throughout the bone from the joints to the diaphyses (Dagan et al., 2004).

2.2.4 Muscles

Nine muscles span the shoulder joint, and four of these muscles, supraspinatus, infraspinatus, teres minor, and subscapularis, the rotator cuff muscles, provide considerable support and reinforce the shoulder joint (Van De Graaff, 2000). The brachium muscles consist of two groups of muscles which cause flexion and extension at the elbow joint. The flexor muscles are the biceps brachii, brachialis, and brachioradialis and the extensor group consists of the triceps brachii and anconeus. The muscles of the forearm can be grouped based on action: supination and pronation of the hand; flexion of the wrist, hand, and fingers; extension of the hand and fingers. Precise finger movements that require abduction and adduction with flexion and extension are caused by small intrinsic muscles of the hand. These muscles are organized into the thenar, hypothenar and intermediate muscle groups. Figure 2 and 3 illustrate the muscles of the arm and forearm, respectively.

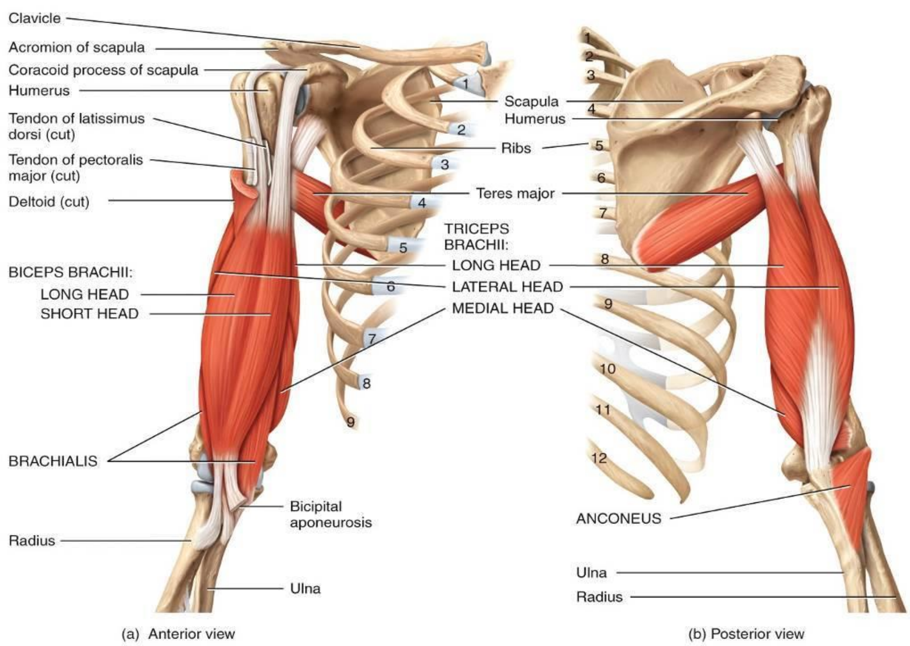

Figure 2: Musculature of the arm (Tortora & Nielsen, 2014).

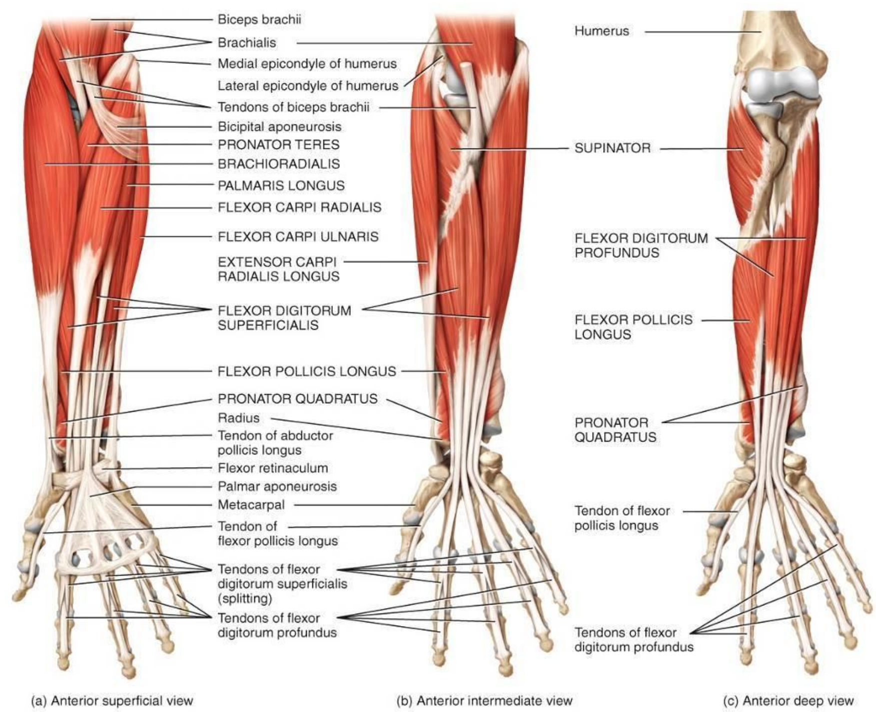

Figure 3: Musculature of the forearm (anterior view) (Tortora & Nielsen, 2014).

2.3 Tissue Behaviour of the Upper Extremity

Muscles behave actively and passively, a unique quality compared to other body tissues. The active component of muscle is regulated by muscle length, contraction velocity, and activation function (Johansson et al., 2000) in order to move and stabilize the body (Van De Graaff, 2000). Muscles amplify impact shock by actively increasing the stiffness of the body's segments (Holmes & Andrews, 2006). Passive muscle force is associated with the connective tissues of the muscles, and increases exponentially as muscle length increases. Muscle lengthening stores elastic energy within the muscle which can be used to reduce the energy costs of movement and postural support (Van Eijden et al., 2002).

The active and passive components of muscle tissue affect impact loading within the body. Schinkel-Ivy et al. (2012) found that the shock wave response of the tibia was affected by the passive soft tissues of the leg through movement of wobbling mass (skin, fat mass and lean/muscle mass) relative to the underlying bone. Their results suggest that the acceleration response differences from an impact were partially caused by the composition of the passive soft tissues. As tissue mass increased peak acceleration and acceleration slope decreased significantly and time to peak acceleration decreased significantly. Muscle activation causes an increase in muscle stiffness, that causes an increase in impact force (Butler et al., 2003; Nigg et al., 1995), and loading rates of ground reaction forces experienced during impact (Butler et al., 2003). An increase in stiffness of the muscles during a forearm impact contributed to a greater peak impact force and a decrease in the time to peak force (Pain & Challis, 2002).

In addition to the increase in muscle stiffness during impact, activation of the muscles causes an increase in the stiffness of the joint(s) that the muscle crosses. Elevated joint stiffness (reduced range of motion) is associated with increased impact forces as well. Milner et al. (2007) found that runners with increased joint stiffness at the knee experienced increased vertical ground reaction forces during the impact phase.

Changing the elbow joint angle at impact has been assessed as a method for reducing the loads applied to the upper extremities during a forward fall (Burkhart & Andrews, 2010). Chou et al. (2001) studied twenty male participants in a simulated fall and compared the energy absorption abilities of the elbow and shoulder during elbow and shoulder flexion at impact. The impact force experienced by the upper extremity was reduced by 68% when participants flexed their elbows, compared to when they were in full extension. Similar reductions in impact forces with flexed elbows were found by DeGoede & Ashton-Miller (2002). Having the elbow flexed during a forward fall acts to dampen and absorb the impact energy, reduces the magnitude of the peak impact force, and delays the peak force temporally (Chou et al., 2001).

2.4 Impact Forces

Impact forces occur during a collision between bodies. The peak amplitude of an impact force occurs after impact has occurred (Nigg et al., 1995). Ground reaction forces (GRF) are equal in magnitude and opposite in direction to the forces that the body exerts on the ground during impact. During a fall with outstretched hands, hand contact forces are characterized by an initial high-frequency oscillation followed by a lower-frequency oscillation as the force wave (shock wave) travels through the forearms, elbows, arms, and shoulders of the upper extremities (Robinovitch & Chiu, 1998).

The total mass of the rigid tissues (bone) of the body has been shown to affect peak vertical impact forces experienced by the body during an impact. Liu & Nigg (2000) found that, as the amount of rigid mass increased within the lower extremity, the impact force peak increased as well. Pain and Challis (2004) reported that the peak vertical GRFs increased as bone mass increased; a 20% increase in bone mass produced a 13% increase in peak vertical GRF.

Oscillations of the soft tissue have been shown to significantly reduce the amplitude of impact-induced shock waves through the leg (Wakeling & Nigg, 2001). The impact force creates a shock wave that propagates through the tissues, which attenuate the amplitude of the shock wave as it travels away from the impact site (Schinkel-Ivy et al., 2012). Nigg et al. (1995) described shock as a transient condition that occurs after a sudden change in applied force which causes a system to deviate from equilibrium from stressed to unstressed tissue. Shock attenuation is described by Nigg et al. (1995) as the reduction in impact force amplitude as it travels through the tissue of the body. The term shock wave has been used throughout the literature by various authors to describe the response of the lower extremity (Derrick et al., 1998; Dufek et al., 2009; Flynn et al., 2004; Hamill et al., 1995; Holmes & Andrews, 2006; Lafortune et al., 1991; Lafortune et al., 1995, 1996; McMahon et al., 1987; Mercer et al., 2002, 2003a, b; Milner et al., 2006; Verbitsky et al., 1998; Voloshin et al., 1998; Voloshin & Wosk, 1982; Whittle, 1999; Wosk & Voloshin, 1981; Schinkel-Ivy, 2010; Schinkel-Ivy et al., 2012), and upper extremity (Burkhart & Andrews, 2010; Hwang et al., 2006; Kim et al, 2006; Greenwald et al., 1998; DeGoede et al., 2002; Lo et al., 2003) to impact.

The peaks of the impact forces can be reduced actively through the adjustment of body kinematics, muscle activity (Cholewicki & McGill, 1995), and joint positions (Hamill et al., 1995), or though passive movement of wobbling mass relative to bone (Hamill et al., 1995; Lafortune et al., 1996). Flynn et al. (2004) and Holmes & Andrews (2006) showed that varying levels of muscle activation in the human leg can change the impact forces that propagate though the leg. It has been suggested in these studies, that higher levels of muscle activation increased muscle stiffness in the leg, which reduced the muscles' ability to dampen the impact shock wave and increased the acceleration of the shock wave through the body. However, the soft tissues of the leg have been considered as a single unit in previous research and the individual contribution of the rigid and soft tissues to shock attenuation has not been researched in any systematic way with living tissue (Schinkel-Ivy, Burkhart, & Andrews, 2012).

2.4.1 Injury

Distal radius fractures represent the most common type of fracture in people under the age of 75 years, and fracture incidence increases with age, representing the fastest growing cause of disability in the elderly (Chiu & Robinovitch, 1998). Forward falls cause approximately 90% of all fractures of the distal radius (Chou et al., 2009). Understanding the biomechanical factors that increase risk of upper extremity injury is important to help create interventions that will reduce these risks (Nevitt & Cummings, 1993).

Forearm fractures are estimated to account for 20% of all reported fractures in the world, with the most common fracture occurring at the distal radius (Johnell & Kanis, 2006). In Europe, fractures result in more disability adjusted life years lost than any other disorder except lung cancer (Johnell & Kanis, 2006). In the developed world, fractures are a significant cause of morbidity and mortality in the older population (>50 years of age) (Johnell & Kanis, 2006). Ninety percent of distal radius fractures occur due to forward falls (Nevitt and Cumming, 1993). Distal radius fractures (e.g., Colles' fractures) are a common injury with older individuals because they are at an increased risk of falling. These injuries can result in long-term problems such as reduced range of motion and pain (Troy & Grabiner, 2007), which can lead to a decrease in the ability to perform daily physical activities. Because distal radius fractures are usually caused by impact from standing height or lower, they are considered low-energy fractures (Troy & Grabiner, 2007).

During a forward fall, the upper limbs are typically extended by people to protect their heads and torsos. This protective reaction brings the hands into contact with the ground, causing a shock wave which travels through the hands and through the forearm tissues to the elbow and arm more proximally (Burkhart & Andrews, 2010). A positive relationship has been found between impact force and injury risk. The tissues of the body experience an increase in stress as the magnitude of impact forces increase. These stresses need to be attenuated to prevent injury (Dufek & Bates, 1991). Duma et al. (2003) used human cadaver arms that were disarticulated mid-humerus and supported by two cables that held the elbow at 90° flexion during axial compression of the wrist. The forearm tissue was kept intact to preserve the load distribution throughout the forearm as the shock travelled toward the humerus. The results showed that, as the impact force increased, the probability of wrist injury increased (Duma et al, 2003).

It has been found that tissue composition influences the risk of injury in the body. Studies have found a positive relationship between tissue mass and risk of injury, including how lower bone mineral content and muscle mass increase the risk of injury from an impact (Bennell, et al., 1996; Kelsey et al., 2007; Loud et al., 2007; Magnusson et al., 2001). Burkhart et al. (2013) found that, as the fat mass to bone mineral content ratio decreased in the leg, the risk of injury decreased, presumably due to increases in the strength of the bone that would come with increased bone mineral content, relative to fat mass. Other factors such as rate of loading, pre-existing conditions, repetition, and past history of injury also affect the risk of injury. It has been suggested that the rate of loading decreases and the risk of injury from an impact decreases as tissue mass of the body segment increases (Schinkel-Ivy, Burkhart, & Andrews, 2012). Pre-existing conditions such as osteoporosis increase the risk of injury related to a fall. A higher susceptibility of bone fracture with osteoporosis is caused by a decrease in trabecular and cortical bone density and by a change in structural properties that occur with osteoporosis (Johnell, Gullberg, Allander, & Kanis, 1992). Repetitive impacts, combined with past history of falls, which could damage or weaken the tissues (i.e., bone), increases the risk of a more serious injury such as a bone fracture (Butler, Crowell, & Davis, 2003 & Milner et al., 2006).

2.5 Biomechanical Modeling

Traditional rigid segment biomechanical models ignore the effects that segment soft tissues have on impact force attenuation (Burkhart, Arthurs, & Andrews, 2008). Rigid segment models oversimplify the human body by only using rigid bodies to represent each segment being modelled. The lack of accurate in-vivo measurements and the limited amount of cadaver data in the literature have made it hard to develop more sophisticated models that more accurately represent the response of the human body to impact (Clarys et al., 1999; Martin et al., 2003). Soft tissues, compared to rigid segments alone, have been shown to influence the impact forces being transmitted through the body (Gittoes, Brewin & Kerwin, 2006; Gruber et al., 1998; Liu and Nigg, 2000; Pain & Challis, 2006; Yue and Mester, 2002). Models that incorporate soft tissues, or wobbling mass, account for the muscle movement in response to an impact such as falling (Schinkel-Ivy et al., 2012). Some authors have compared the outputs of rigid mass models to those of wobbling mass models and the outputs of energy, impact forces, joint forces, and joint torques have been shown to be dramatically different (Gruber et al., 1998; Pain & Challis, 2001, 2006). Wobbling mass models are useful to show the dissipation of energy during impact and the reduction in external impact loads, joint forces, and joint torques (Pain & Challis, 2001, 2004, 2006; Gittoes et al., 2006; Gruber et al.,1998; Liu & Nigg, 2000). Cole et al. (1996) and Gittoes et al. (2006) have shown that wobbling mass models were a better predictor of experimental kinematic measurements than rigid mass models. Liu & Nigg (2000) concluded that rigid mass models are beneficial for research of slow, quasi-static movements but not for impact studies. Overall, the results of modeling studies show the importance of incorporating wobbling mass into impact research, because accounting for soft tissue motion provides more accurate and reliable data of segmental responses to impact events.

2.6 Shock Transmission and Attenuation

Shock waves generated from impact during a fall onto the upper extremities travel proximally from the hand towards other anatomical structures including the wrist, forearm, elbow, arm, and shoulder (Burkhart & Andrews, 2010). Shock wave transmission is characterized by the non-linear properties of the tissues (bone, muscle, skin, adipose, connective tissue) from which the segments and joints are comprised (Lafortune et al., 1996).

Accelerometers have been used in research by Burkhart & Andrews (2010) to measure shock of the upper extremity following an impact. They have also been used in numerous studies to measure shock during impact in the lower extremity (Lafortune, Henning, & Valian, 1995; Radin et al., 1973; Davis et al., 2004; Milner et al., 2006; Schinkel-Ivy, 2010; Schinkel-Ivy et al., 2012). Wosk & Voloshin (1981) and Radin et al. (1973) used accelerometers on the lower extremity and trunk and found a strong relationship between peak magnitude of the transient shock wave and the risk of injury. Wosk & Voloshin (1981) quantified shock attenuation based on the magnitude of the PA, TPA and AS at the tibial tuberosity, medial femoral condyle, and the forehead during walking. They found that 30% of the PA was attenuated across the knee joint and 40% of the remaining shock wave was attenuated by the more proximal body regions, for an overall attenuation of 70% (Wosk & Voloshin, 1981). Wosk & Voloshin (1981) and Schinkel-Ivy (2010, 2012) examined how the body attenuates impact shock using accelerometers by quantifying PA, TPA, and AS in the leg. Similarly, Burkhart & Andrews (2010) studied how the forearm attenuated shock from a forward fall impact using accelerometers at the hand and elbow. The amount of shock wave attenuation by the body has a positive relationship with the magnitude of the shock resulting from impact to the body (Hamill et al., 1995). Derrick et al. (1998) and Mercer et al. (2003)

showed that, as impact forces increased, shock attenuation by the tissues of the body also increased.

2.6.1 Shock Propagation Through the Body and Effects of Soft Tissue

The speed of shock wave propagation along the longitudinal axis of long bones in cadavers is approximately 3200 m/s (Chu et al., 1986; Pelker & Saha, 1983). Pain & Challis (2002) showed that after a forearm impact, the system (forearm segment) returned to equilibrium in 200-250 ms. Similar results for the time it takes to return the segment to equilibrium were reported by Lafortune et al. (1996) for the lower extremity. However, in in-vivo studies, the speed of propagation is reduced by the movement of soft tissues relative to the underlying bone (Dufek et al., 2009). A reduction in the speed of propagation occurs because of the response of the tissue to the applied load during impact, which reduces the acceleration of the shock wave through the tissue (Challis & Pain, 2008). The impact force is propagated proximally, causing deformation of the soft tissue in a wave-like motion, which occurs because of the unbalanced forces acting on the distal end of the extremity (Nigg et al., 1995).

Pain & Challis (2002) measured marker motion from proximal and distal points on the soft tissue of the upper extremity and determined the shock wave propagation velocity though the soft tissue to be 37 m/s; approximately 100 times less than the velocity through cadaver bone (Chu et al., 1986; Pelker & Saha, 1983). These shock wave propagation velocities show that there is a high frequency component of the wave that travels through bone and a low frequency component that travels through soft tissue as a result of impact (Challis & Pain, 2008). As the shock wave travels proximally after impact, a portion of the wave is reflected back distally which results in non-uniformity of the stress on the tissues (Nigg et al., 1995).

Schinkel-Ivy et al. (2012) studied 18 participants using a pendulum device for impact testing to investigate the relative effects of body composition and individual leg tissue masses on the acceleration response of the tibia following impact. The results of the study show that, as wobbling mass to bone mass ratio increased, the acceleration response of the tibia becomes smaller in magnitude and longer in duration. This suggests that the impact force attenuation in people with a greater wobbling mass to bone mass ratio is higher than people with a lower wobbling to bone mass ratio (Schinkel-Ivy et al., 2012).

2.6.2 Passive versus Active Shock Attenuation

Shock attenuation can occur in the body passively or actively. Shock is passively attenuated by structures of the body such as cartilage and wobbling mass (muscle, skin and adipose tissue, ligaments, and tendons) (Holmes & Andrews, 2006). Passive attenuation is increased as the shock wave travels through tendons, ligaments and other structures such as the interosseous membrane (Williams et al., 2001).

Active shock attenuation is primarily caused by the activation of the muscles of the body, due to muscles' capacity for deformation in response to impact loading, and their ability to adjust the amount of shock attenuation by varying activation levels (Derrik et al., 1995). As muscle activation increases, so does the stiffness of the muscle, which reduces its shock attenuation capabilities (Mercer et al., 2003b).

Holmes & Andrews (2006) studied the effects of increased muscle activation on the transmission of shock through the leg, using a human pendulum to perform controlled impacts. Acceleration waveforms measured medial to the tibial tuberosity indicated that increased muscle activation resulted in increased PA and AS and a decrease in TPA leading to a reduction in shock attenuation (Holmes & Andrews, 2006).

2.7 Methods of Upper Extremity Impact Testing

In-vivo research utilizes living participants. Certain variables such as muscle contractions of the hand, forearm, and arm affect the impact response during a fall and using living participants allows these variables to be analyzed (Chiu & Robinovitch, 1998). However, each participant can perform the action differently, which can have erroneous effects on results (Chou et al. 2009). Use of living participants restricts fall experiments to safe fall heights in order to reduce the risk of injury to the participants. Researchers have to use models or in-vitro techniques to predict impact forces at actual falling heights (Chiu & Robinovitch, 1998). In-vitro research simulates an impact such as a fall and uses variables (force, acceleration, and velocity) that are similar to a fall impact that cannot be performed by living participants due to risk of injury. In order to investigate the effect that each tissue layer has on shock attenuation from a forward fall, in-vitro testing techniques must be utilized. Described below are a few in-vivo and in-vitro methods that have been used to date to study impacts from simulated forward falls on the upper extremities.

2.7.1 In-Vivo Research Methods

Burkhart & Andrews (2010) used a novel seated human pendulum device (Figure 4) that was designed to simulate the flight phase of a forward fall, to deliver bilateral impacts to the upper extremities. Impact reaction force was measured using a force platform, and impact velocity was monitored by a velocity transducer. Accelerations of the distal radius and proximal ulna of the right forearm of the participants were measured to determine the transient impact force effects. Accelerations were measured along the long axis of the forearm (axial direction), and at right angles to the axial direction. Three acceleration dependent variables were measured from the data (PA, TPA, and AS) (Burkhart & Andrews, 2010). A limitation of the study was that participants were in a seated position for impact, which does not represent the positioning of the body during a natural forward fall.

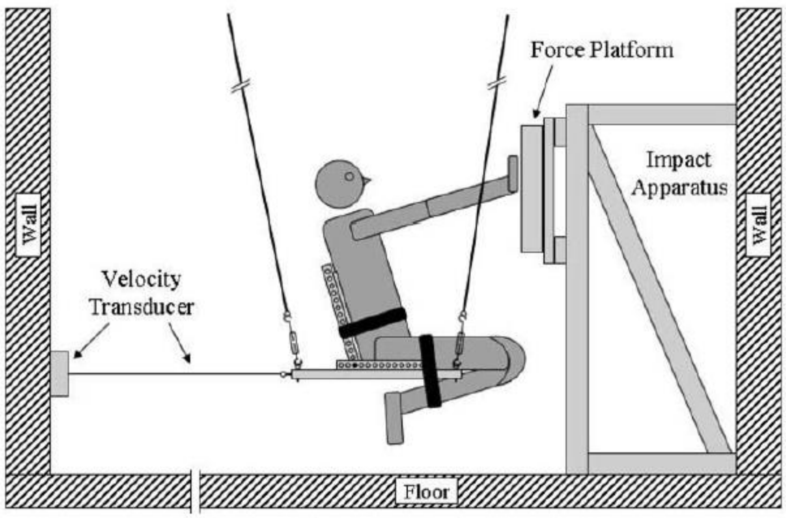

Figure 4: An illustration of the pendulum apparatus used by Burkhart & Andrews (2010) to simulate a forward fall impact.

Chou et al. (2001) simulated forward falls to compare the energy absorption abilities of the elbow and shoulder joints at impact, comparing relaxed and flexed muscles. In a similar study by Chou et al. (2009), participants performed three one-armed falls, each with a different forearm rotational posture of the hand. With help from a suspension system, participants were dropped from a height of 5 cm onto a force platform with their knees in contact with the ground at all times. The angle between the trunk and arm was maintained at 60° to simulate a forward fall (Figure 5). A set of eleven reflective markers were placed on selected anatomic landmarks on each participant to monitor body position, and impact force was recorded using a force platform. Three postures of the forearm were used in this study at impact: forearm externally rotated 45º, internally rotated 45º, and in a neutral position (not rotated), at impact (Chou et al., 2009). A limitation of the study was that only forces were collected at impact using a three-segment model rather than using accelerometers to determine the transient impact forces in all three axes.

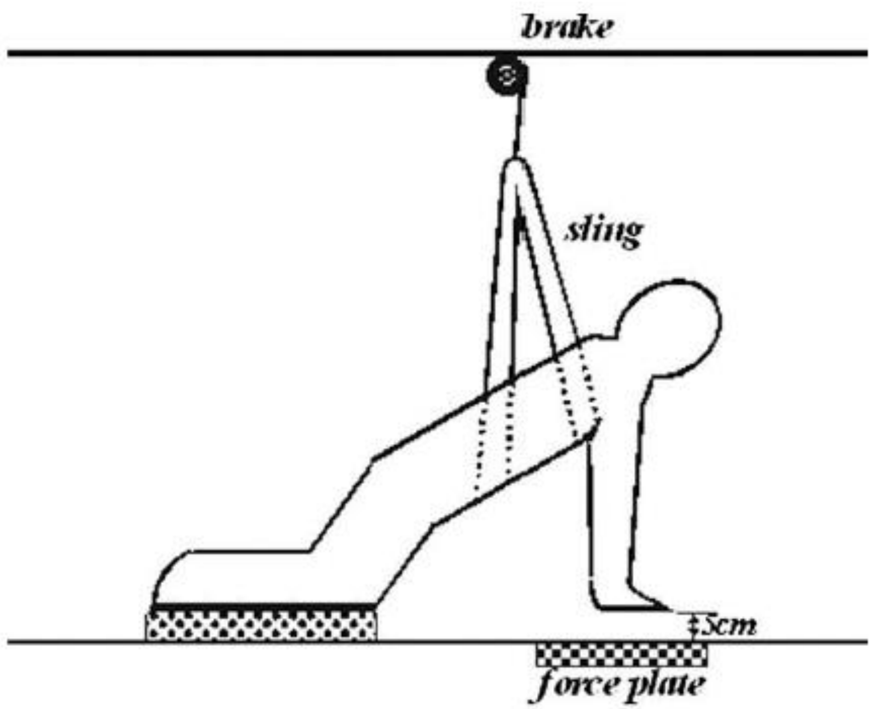

Figure 5: An illustration of the fall apparatus used by Chou et al. (2009).

Burkhart, Andrews & Clarke (2012) used the Propelled Upper Limb fall Arrest Impact System (PULARIS) (Figure 6) to simulate the impact phase of a forward fall on the upper extremities of healthy, young participants. The advantage of using PULARIS is that the impact velocity and body position of the participants are similar to those experienced during a forward fall. Participants were propelled at approximately 1.0 m/s towards two force platforms mounted to the floor and then quick released 150ms prior to the force platforms to allow for a brief flight phase in advance of hand impact. Participants performed three repetitions of three different fall types (straight arm, self-selected and bent-arm) at impacts at two different heights (0.05 m and 0.10 m). Peak vertical (Fz), medio-lateral (Fx), and anterior-posterior (Fy) impact forces were determined from the force-time curves for both left and right forearms.

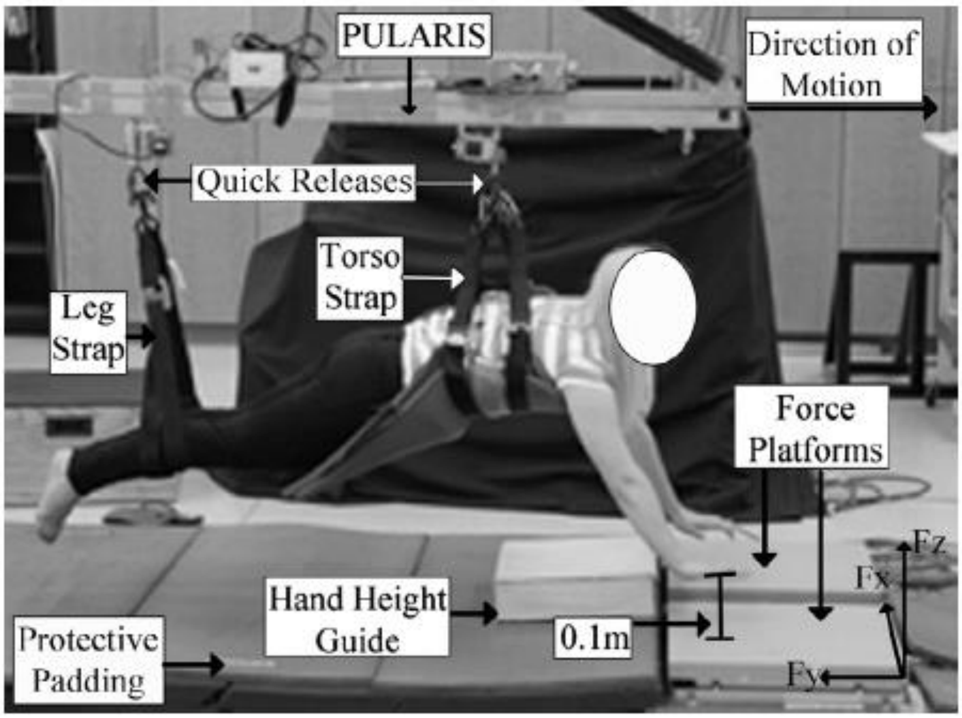

Figure 6: PULARIS setup for simulating a forward fall (Burkhart & Andrews, 2012).

Burkhart & Andrews (2013) used the PULARIS in a similar study to research the effects of different fall types and fall heights on the kinematics, kinetics and muscle activation levels of the upper extremity during a simulated fall. Two tri-axial accelerometers were used on the distal radius and proximal ulna to quantify the impact induced shock wave, and muscle activation levels were collected from six muscles of the upper extremity to quantify muscle response to impact. Each participant performed three repetitions of the same three fall types as reported previously (Burkhart, Andrews &

Clarke, 2012). An advantage of using PULARIS in both studies described here is that it allowed participants to experience different fall types that more accurately represent the multi-directional motion of an actual forward fall. A limitation of both PULARIS studies is the relatively close proximity of the hands to the force platforms at the time of fall initiation. These low fall heights were chosen to ensure participant safety, compared to falling from standing height.

2.7.2 In-Vitro Research Methods

Quenneville et al. (2010) developed and utilized a projectile system to test applied impulse loads over a large range of magnitudes and velocities to cadaveric, synthetic, and anthropomorphic test device tibias (Figure 7). The study showed that the projectile system allowed experimental tests covering a variety of impulse levels to be conducted in a controlled environment. Another strength of this type of approach is that any injury events that occur can be directly related to the characteristics of the applied impact loads (Quenneville et al., 2010).

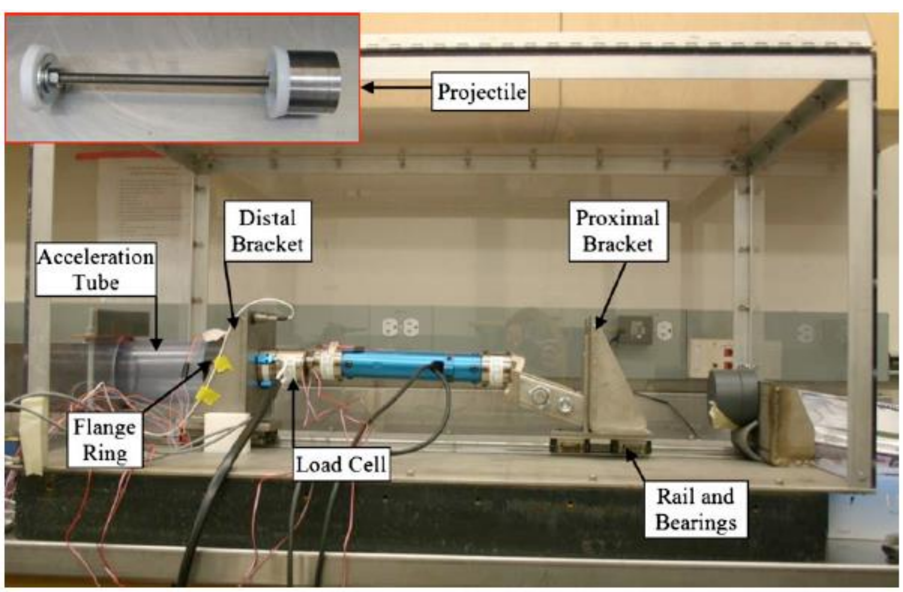

Figure 7: Projectile system used in the Quenneville et al. (2010) study.

The projectile system initially reported by Quenneville et al. (2010) was modified by Burkhart et al. (2011, 2013) to test forearm and hand cadaver specimens during impacts consistent with forward falls (Figure 8). A rail and bracket system was added inside the projectile system to allow weights to be added to the specimen to simulate effective body mass during impact. The bracket system allowed modification of the angle of the forearm relative to the force platform.

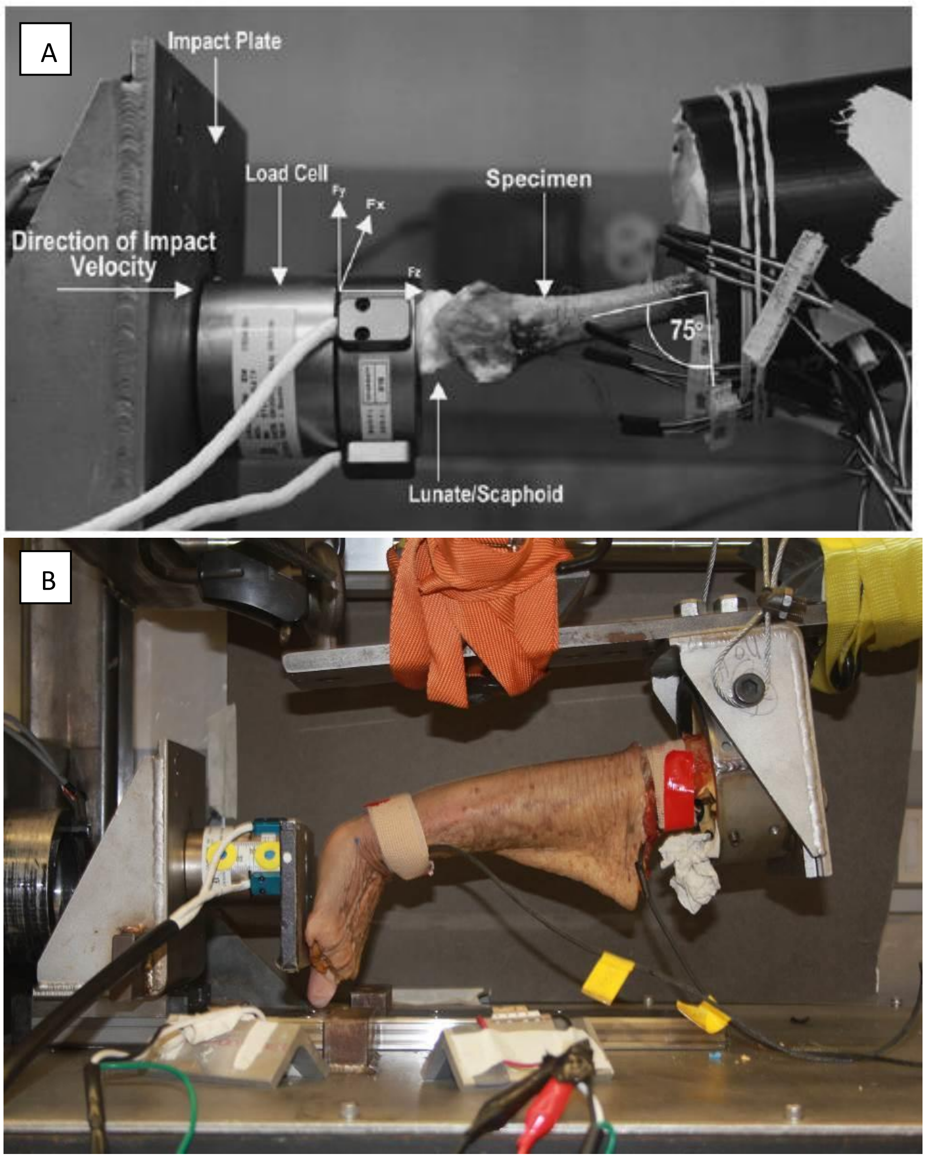

Figure 8: (A) Modified upper extremity projectile system in Burkhart et al. (2011). (B) Further modified upper extremity projectile system used in the current study.

2.8 Summary of Literature Review

The above review of literature provides an overview of some of the research that has been conducted on the effects of rigid and soft tissues of the body during impacts of the lower and upper extremities. Considerable work has been conducted to document the effects of soft tissues on impact shock attenuation for the leg during activities such as running and jumping, and some comparable research that has been performed for the upper extremity during simulated falls. However, there is no known research that compares the relative effects of different forearm tissues (skin and adipose, muscle, and interosseous membrane) on shock attenuation during a simulated forward fall. Quantifying the shock attenuation capabilities of the different tissues of the forearm using a versatile in-vitro method may improve our understanding of how injuries to the upper extremities occur, and ultimately help to improve fall-related injury prevention strategies.

CHAPTER 3

METHODS

This thesis is a secondary analysis of data collected in November 2013 in the Department of Mechanical and Materials Engineering at Western University.

3.1 Specimens

Twelve intact human cadaveric upper extremity specimens, with no previous physical trauma or bone disease specific to the upper extremity (Life Legacy Foundation, Arizona, US), were tested (Table 1). The three female specimens fractured during testing and were therefore not used in the analyses described in this thesis (see below).

Table 1: Age at death, sex and major diseases/conditions identified for each specimen.

Specimen Number	**Age at Death (years)**	**Sex**	**Major Diseases/Conditions Identified**
1	91	F	dementia
2	89	F	pneumonia, arthritis (lower back), metastatic liver cancer
3	81	M	COPD, congestive heart failure, hypertension, dementia, osteoporosis (lower back)
4	78	M	cardio pulmonary disease, heart attack (x 5), diabetes, stroke, COPD
5	80	M	congestive heart failure, stroke, pneumonia
6	82	M	congestive heart failure, dementia, arthritis (bilateral knees, shoulder), pneumonia
7	71	M	lung cancer, hypertension, COPD
8	81	M	dementia, hypertension, diabetes
9	87	M	dementia, coronary artery disease, hypertension, stroke, Alzheimer’s
10	81	M	degenerative disease of basal ganglia, diabetes, arthritis (bilateral shoulders, right elbow)
11	74	M	lung cancer, hypertension, heart attack, insulin-dependent diabetes mellitus,
12	78	F	congestive heart failure, heart attack, osteoarthritis (hands)

3.1.1 Specimen Preparation

The specimens were stored in a standard upright deep freezer and thawed the day before being tested. The elbow joint of each upper extremity was disarticulated with a scalpel and approximately 5 cm of tissue was cut away from the proximal forearm (overlaying the proximal radius and ulna) with a scalpel to enable adequate potting of the forearm and hand specimen into the potting cup (see below). The fingers were removed with a scalpel at the metacarpo-phalangeal joints to facilitate specimen placement within the custom impact apparatus (Figure 9). Movement of the specimens during testing could have been restricted by the fingers rubbing against the floor of the apparatus if they were left intact. Three Philips head, 7.5 cm long screws were placed through the head of the radius and proximal ulna (following pilot holes made using a hand drill) to keep the forearm secured in a fully pronated posture during testing. The screws were needed to enable proper potting, which involved adhering dental cement (Denstone Dental Cement; Modern Materials, Heraeus Holding GmbH, Germany) to the proximal 5 cm of the radius and ulna. The proximal ends were potted in 8.89 cm (3.5 inch) diameter PVC tubing, at a 90° angle with respect to the vertical plane with no frontal plane tilt (Staebler et al., 1999). The angle of the specimen relative to the projectile system once mounted was 75º, which is consistent with the position of the forearm at impact with the ground during an actual forward fall (Greenwald et al., 1998). A compression sleeve (Surgilast, Glenwood Laboratories Canada, Oakville, Ontario) was used to keep the soft tissues from sagging due to gravity (compare Figure 9 and Figure 10).

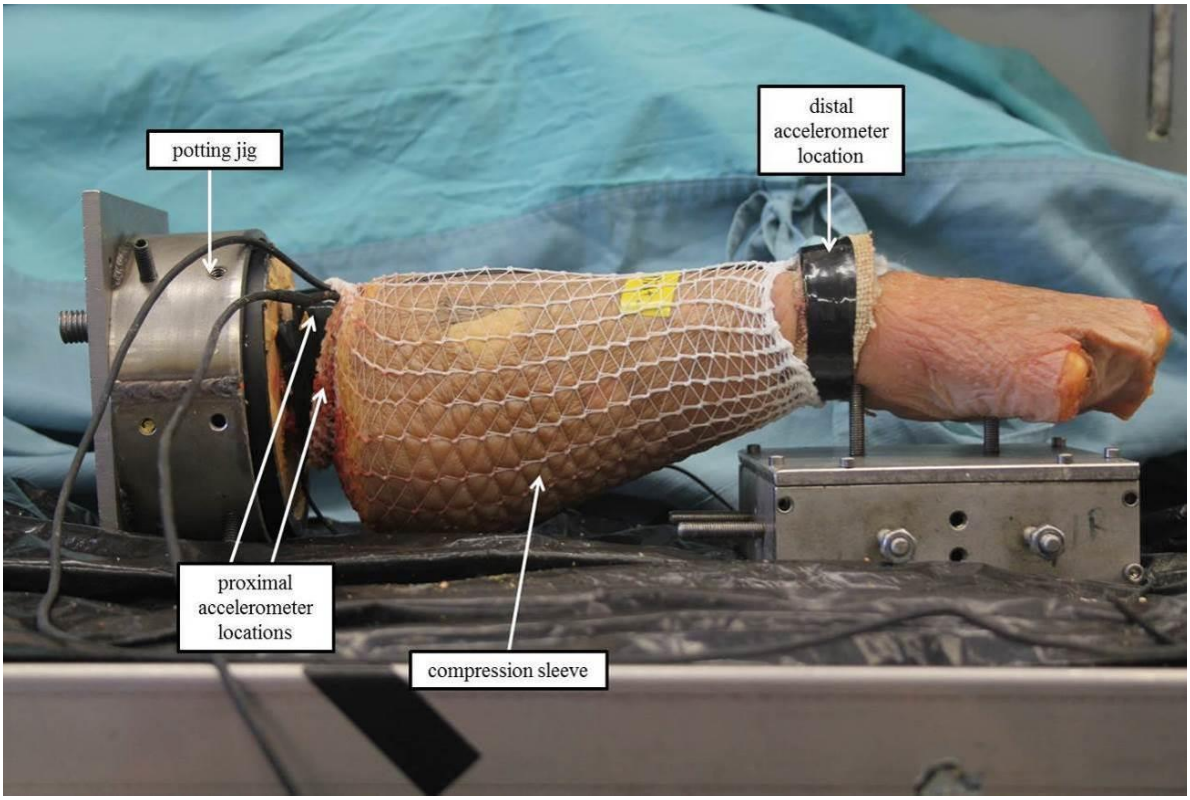

Figure 9: Cadaver forearm and hand specimen potted, with accelerometers attached and compression sleeve in place.

The potted end of the specimen was attached to a bracket, which was on two linear rails at the top of the testing chamber. This allowed the specimen to be placed in the correct orientation against the impact plate (Figure 10). Two clamps on the rails allowed the specimen to move approximately 2-3 cm after impact. The palm of each specimen's hand was placed against a metal plate to allow the wrist extension angle to be approximately 45°, which mimics the angle during a forward fall by a living person (Troy & Grabiner, 2007). After each impact the specimen was moved back to the starting position where the palm was touching the impact plate at wrist extension angle at approximately 45º allowing for relatively the same starting position for every impact.

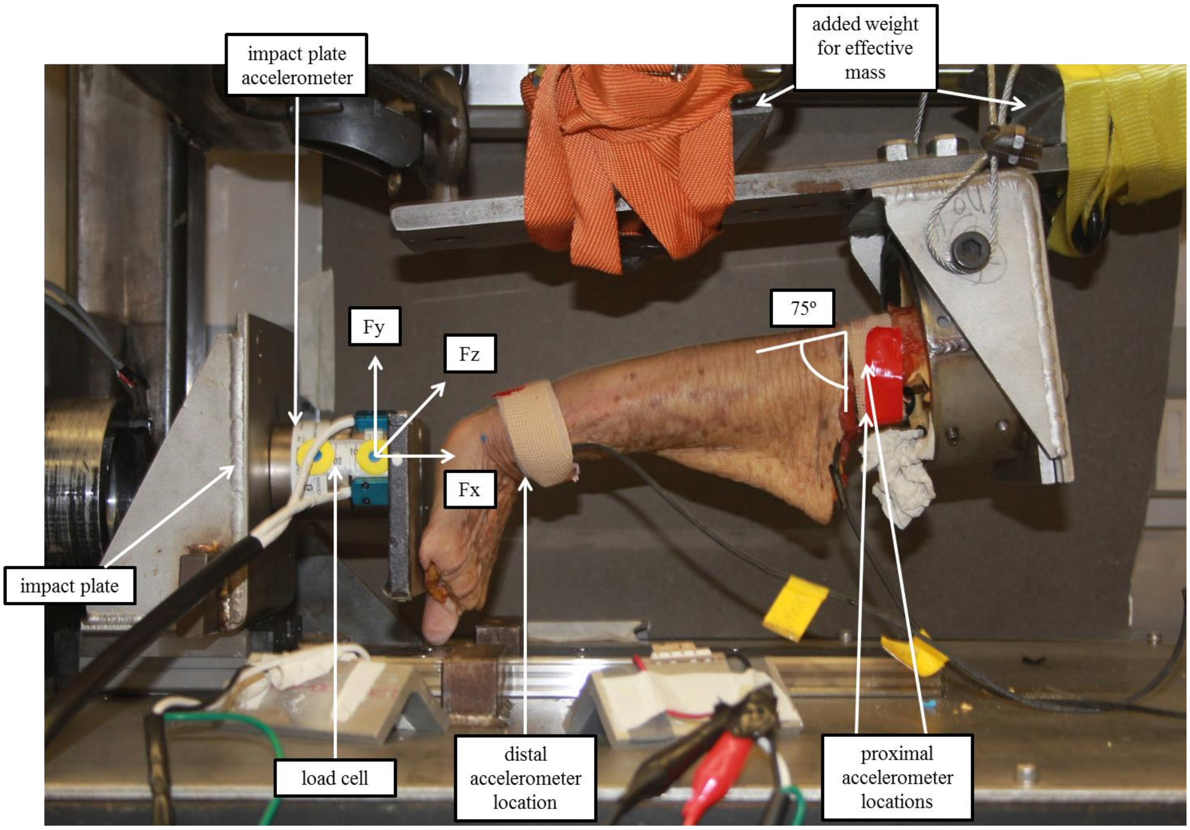

Figure 10: Cadaver forearm and hand specimen placed in the custom impact system. The compression sleeve was not in place for this picture.

The effective mass of the potted specimen, bracket system and added weights was calculated to be 30% of the body mass of the participant, from which the specimen came. Pre-weighed metal blocks were secured to the metal support plate above the specimen to obtain the effective mass for each specimen (Figure 10), which was consistent with the effective mass of a person during a forward fall (Greenwald et al., 1998).

3.2 Instrumentation

Instrumented specimens were placed within a custom designed pneumatic impact system, which was adapted from Burkhart et al. (2011) and Quenneville et al. (2010) (Figure 10). A 6.8 kg projectile was propelled by pressurized air through an acceleration tube (end of tube is visible just to the left of the impact plate in Figure 10). The air pressure was set to 5.8 psi by an input voltage which produced a projectile velocity of 3.0-3.4 m/s, and the release mechanism was activated by a fast-acting solenoid valve (VXR2380, SMC Corporation, Tokyo, Japan). Prior to testing, a pressure-velocity positive relationship was calculated to control the projectile velocity. The velocity used was proportionate to that experienced during a standing forward fall which should not cause a fracture (Chiu & Robinovitch, 1998).

The metal impact plate was attached to a 6 degree of freedom load cell (Denton Femur load cell, model #1914A, Robert A. Denton Inc., Rochester Hills MI; natural frequency 6 kHz) to measure impact force, and optical sensors (TCRT100 Vishay semiconductors, Malvern PA) attached to the impact plate were used to measure velocity (Burkhart et al., 2011).

Three tri-axial accelerometers (Freescale Semi-conductor, Inc., Ottawa, ON, Canada; range of ± 100 G) were glued (M-bond 200; Vishay Micro-Measurements) to the cadaver specimens, in specific locations relative to key anatomical landmarks. Two accelerometers were glued on the bone of the flat surface of the proximal ulna, just distal to the ulnar tuberosity, and on the radius, just distal to the neck (Figure 11). One accelerometer was glued on the skin, dorsal and proximal to the radial styloid (Figure 11); a location previously utilized by Burkhart et al. (2011). The x axis of the accelerometers was orientated parallel to the long axis of the radius and ulna. The y axis was orientated in the medio-lateral direction, and the z axis was orientated antero-posteriorly. Accelerometers were used to measure the response of the radius and ulna following impact because the results obtained from them have been shown to compare favourably to strain gauges, and accelerometers are more economical than strain gauges because they are reusable (Burkhart & Andrews, 2010). Force and acceleration data were collected at 15 kHz (National Instruments NI-PXI 1050, and SCXI 1010) using a custom LabView (LabView 2008, National Instruments, Austin, TX) data collection program (Burkhart et al., 2013).

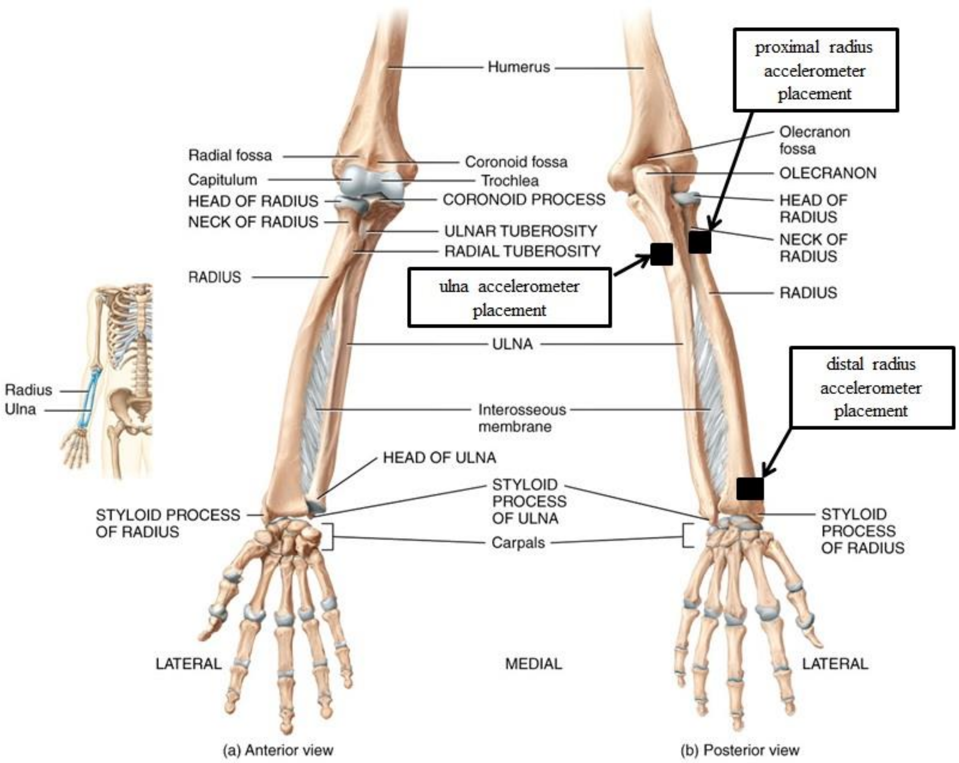

Figure 11: Accelerometer attachment sites on the radius and ulna (Adapted from Tortora & Nielsen, 2014).

3.3 Procedure

Each specimen was impacted a total of eight times at a relative sub-fracture tissue damage level (<2200N; average fracture force (Chui & Robinovitch, 1998; Duma et al., 2003; Burkhart et al, 2013)), similar to what would be experienced during a forward fall. Two impacts were delivered to the specimens in each of four different states: fully intact; after skin and fat tissue had been removed; after the muscle tissue had been removed; and after the interosseous membrane had been cut. After each set of two impacts, each specimen was removed from the impact apparatus. Once removed, the compression sleeve was taken off in order to allow the specimens to be prepared for the next set of impacts. Dissections were carefully carried out using a scalpel. All tissues that were removed between impact sets were weighed and then put aside for the future cremation ceremony. The compression sleeve was used until the muscles were removed. Prior to the final set of two impacts, the interosseous membrane was cut with a scalpel while the specimen remained in the apparatus. Between each set of two impacts, the specimens were carefully evaluated visually. The nine intact specimens (all male) showed no signs of damage, cracks or fractures throughout the testing procedure.

3.4 Data Analysis

The time histories from the distal radial x and proximal ulnar x accelerometers were used to determine the dependent forearm response variable peak acceleration (PA) (Figure 12) (Schinkel-Ivy, 2010; Duquette & Andrews, 2010; Flynn et al., 2004; Holmes & Andrews, 2006). The acceleration data from the accelerometers were filtered using a dual-pass Butterworth low pass filter, the cutoff frequency of 400 Hz which was determined by residual analysis (Winter, 2005). The PA was determined for each acceleration curve to quantify the differences in shock attenuation between tissues. Shock attenuation was calculated using Equation 1:

$$SA = [1-(a_{proximal} / a_{distal})] \times 100 \qquad [1]$$

where SA represents shock attenuation, $a_{proximal}$ represents the PA at the two proximal accelerometers (proximal radius and ulna which were compared separately), and a_{distal} represents the PA at the distal accelerometer (distal radius) (Dufek et al., 2009).

When the magnitude of the proximal PA was less than the distal PA, the SA was positive, showing that the acceleration was attenuated. When the magnitude of the proximal PA was greater than the distal PA, the SA was negative, representing an amplification of the acceleration as the shock wave travelled proximally.

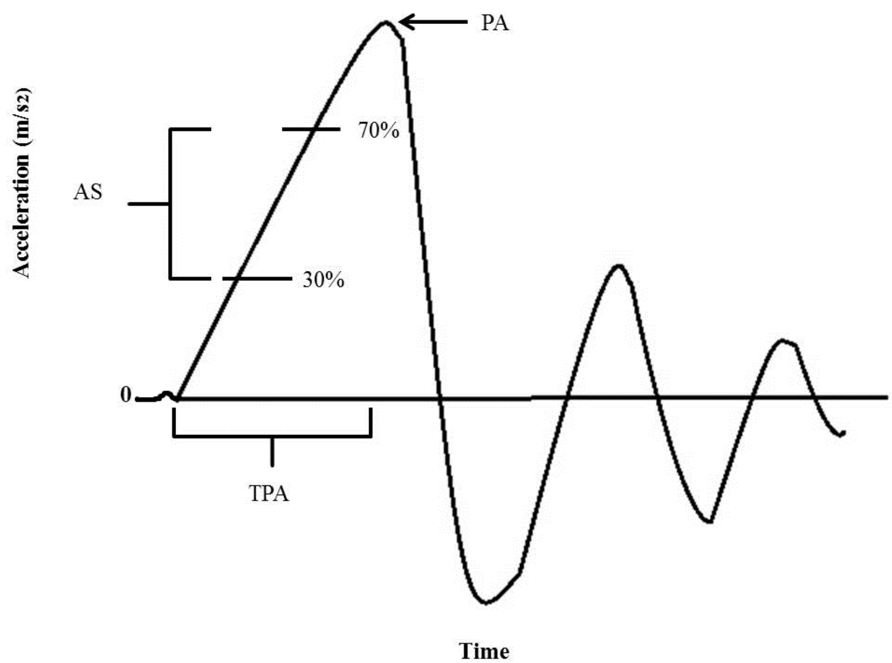

Figure 12: Schematic diagram of an acceleration waveform from an impact of the forearm showing how the dependent forearm response variables were defined. PA: peak acceleration; TPA: time to peak acceleration; AS: acceleration slope. Note that PA was the only one of these variables evaluated in this thesis (see Results).

3.5 Statistical Analysis

The dependent variables for the study were shock attenuation (SA) and peak acceleration (PA). The within-subject factors include: Layers (4) - intact, skin/adipose, muscle, and interosseous membrane; Accelerometers (2) - radius to ulna. Each variable and factor were analyzed and compared along the x axis.

A paired samples t-test was conducted for each dependent variable to see if there was a significant effect of repetition (i.e., impact 1 vs. impact 2). If there was no significant difference in the dependent variables as a function of repetition, the mean value of the two trials was taken for further analysis.

Hypothesis 1: If the SA values are positive then the shock wave was attenuated as it moved from the distal end of the forearm to the proximal end of the forearm. This was determined using Equation [1]. A repeated measures (2 x 4: Accelerometers x Layers) analysis of variance (ANOVA) was performed for the dependent variable PA to determine if there are any significant effects of Accelerometers and Layers on shock attenuation.

Hypothesis 2 & 3: A repeated measures (2 x 4: Accelerometers x Layers) analysis of variance (ANOVA) was performed for the dependent variable SA to determine if there are any significant effects of Accelerometers and Layers on shock attenuation.

All statistical analyses were performed using SPSS 22.0 (IBM SPSS statistics, IBM Corporation, Somers NY) with alpha set at 0.05. Bonferroni Post hoc analyses were conducted as needed on statistically significant main effects or interactions. The system missing values tool in SPSS was used to account for missing data in the ANOVAs.

CHAPTER 4

RESULTS

4.1 Limitations with Data Collection

After close inspection of the data, it was clear that there were a couple of issues with it that were not caught during data collection. These issues would ultimately prove to limit the analyses that could be completed with any confidence. Consequently, only the analysis of peak acceleration and shock attenuation between the distal radial and proximal ulnar accelerometers in the x direction could be addressed. Additional data collection could not occur because of the high cost of the specimens and issues with data collection (some accelerometer channels, impact forces) that could not be repaired in a reasonable time frame. Specifically, the proximal radius accelerometer data could not be used for analysis as there appears to have been malfunctioning channels that were not caught during data collection. The z axis channel for the distal radius accelerometer was unresponsive for all trials, registering only baseline noise. Similar issues with large amounts of noise existed for data from the proximal ulnar and radial accelerometers in the z and y directions, and x direction, respectively. Consequently, these data could not be used in any analysis.

4.2 Peak Acceleration

The mean peak acceleration for impact 1($\bar{x}$ = 26.6 (12.8) g) was not statistically different than the mean value for impact 2 ($\bar{x}$ = 30.0 (11.9) g) (p = 0.44, F = 0.77, df = 1) for all specimens tested. As a result, the mean value of the two impacts was used for further analysis. In general, the proximal accelerations were greater in magnitude than the distal peak accelerations for all specimens (Figures 13, 14). The peak accelerations for all conditions and specimens are illustrated in Figure 14. A significant main effect of accelerometer (distal radius vs. proximal ulna) was found for peak acceleration in the x direction ($p = 0.003$, $F = 24.63$, $df = 1$) (Figure 15). As the shock wave travelled from the distal radius to the proximal ulna, the mean peak acceleration for all specimens decreased from 42.8 (19.0) g to 12.3 (4.3) g. There were no other significant main effects or interactions for peak acceleration.

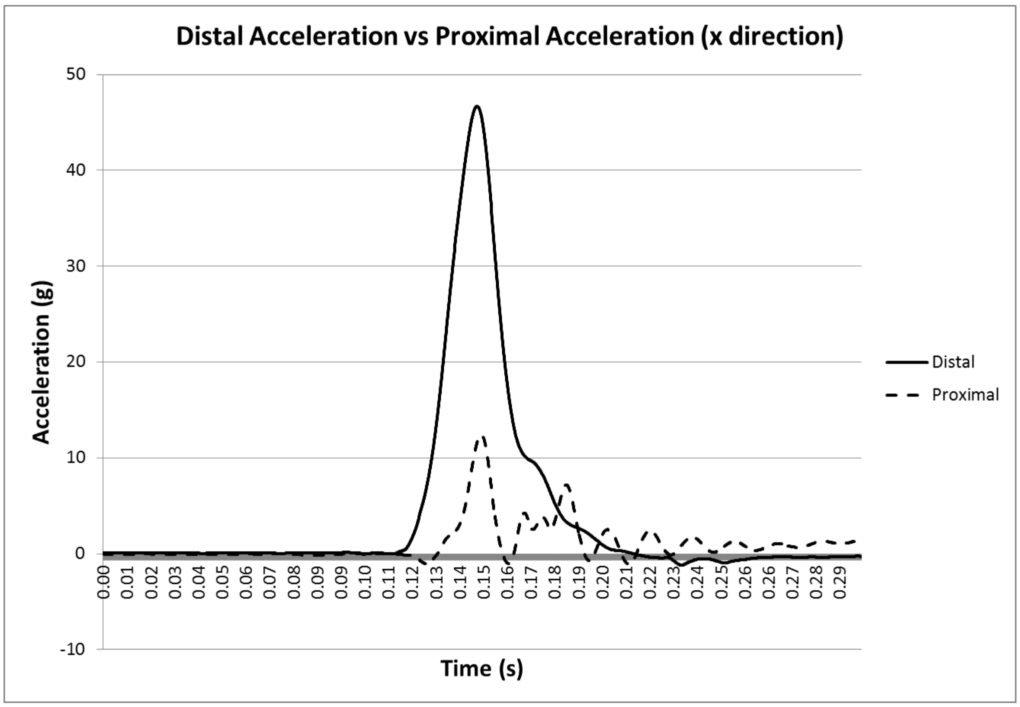

Figure 13: A sample of filtered distal (radius) and proximal (ulna) acceleration waveforms (x direction).

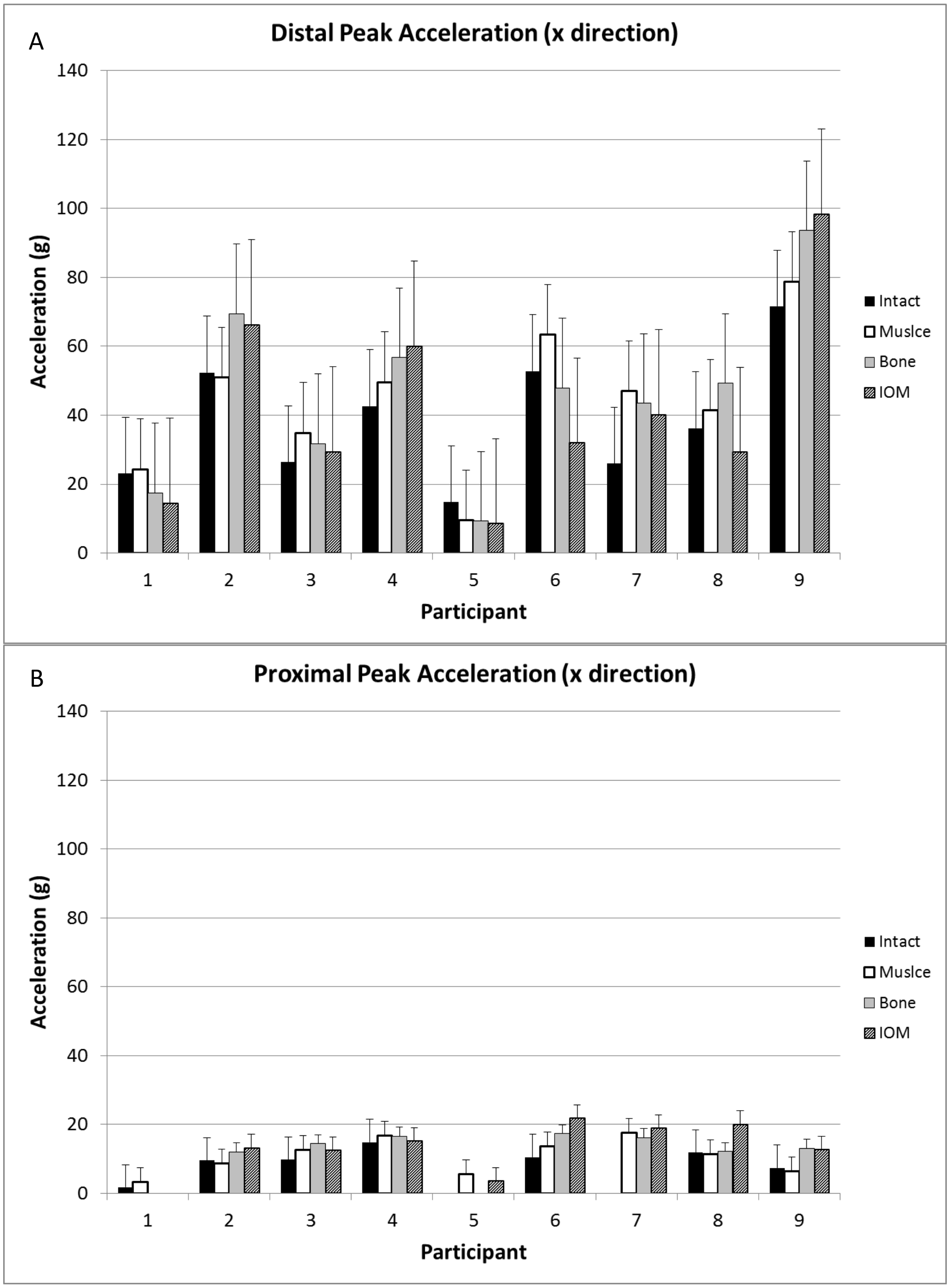

Figure 14: Comparison of distal (A) and proximal (B) peak accelerations in the x direction for all participants. Data from the IOM and Bone trial for participant 1 were unusable.

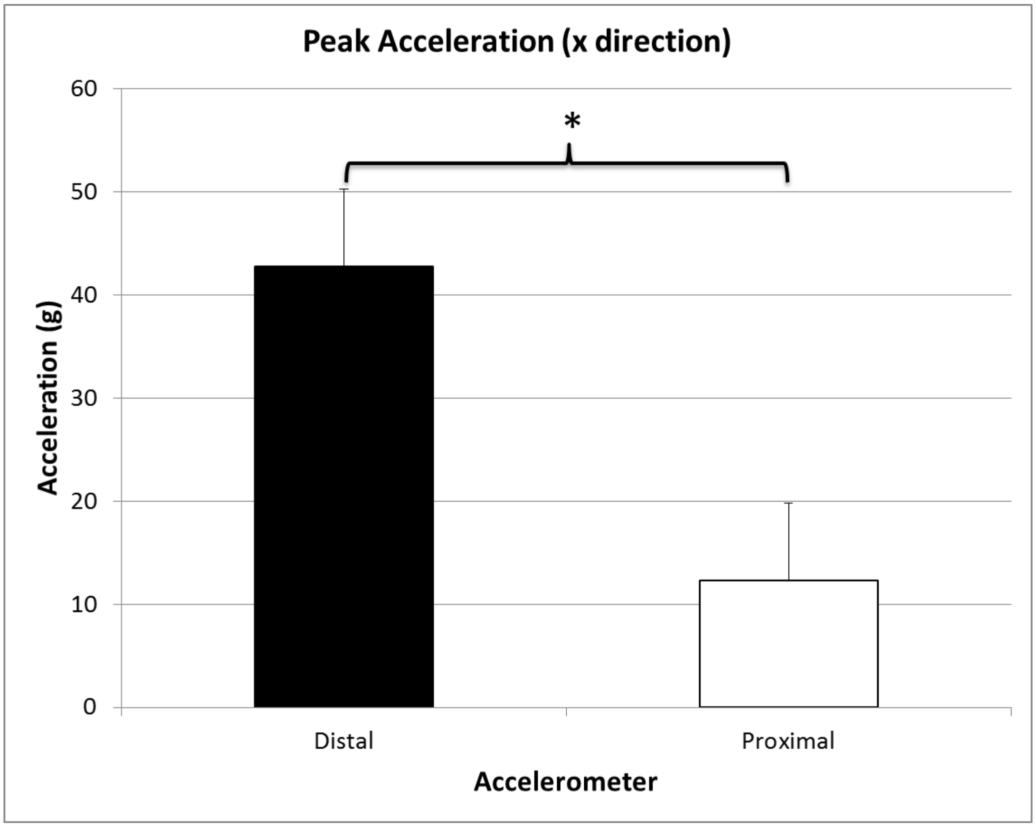

Figure 15: Mean (SD) peak acceleration values between the distal radius and proximal ulna. *Significantly different between accelerometers in the x direction ($p < 0.003$).

4.3 Shock Attenuation

Figure 16 illustrates the shock attenuation for all specimens and the different tissue layer conditions. There was no main effect of Layer ($p = 0.4$, $F = 1.19$, $df = 3$) on shock attenuation (Figure 17). Shock attenuation of the intact condition was 78.1%, which decreased on average 8.9%, 7.5% and 21.9% when the skin and adipose tissue were removed, muscle tissue was removed, and IOM cut, respectively.

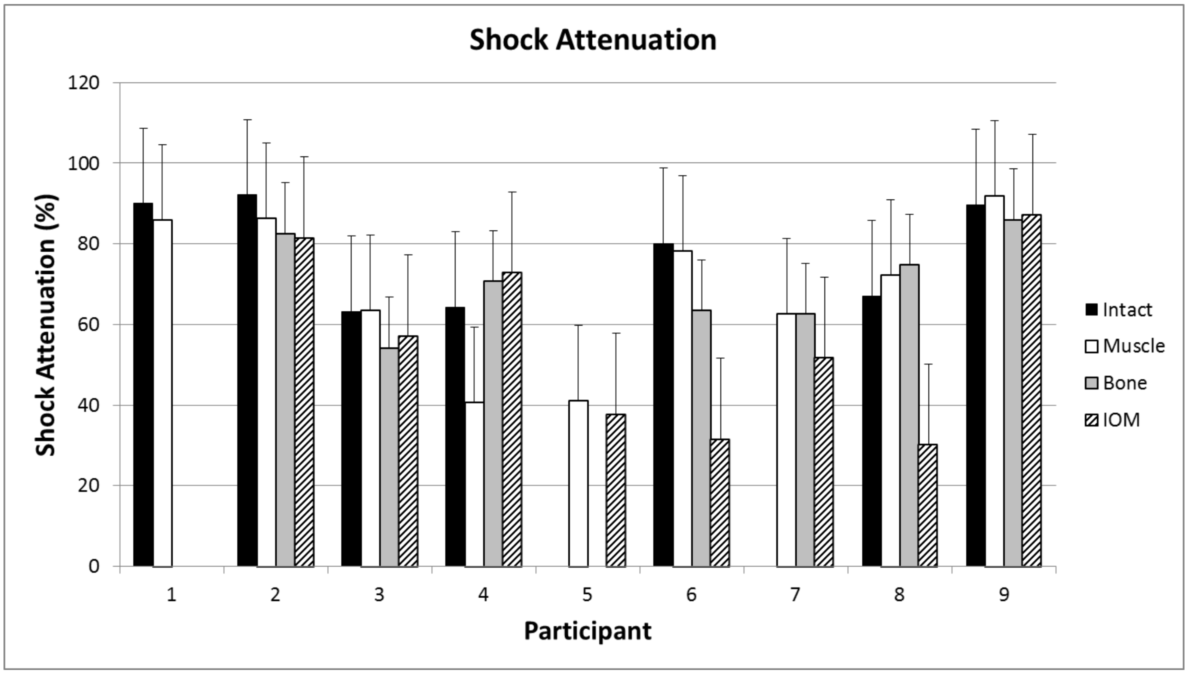

Figure 16: Mean (SD) shock attenuation values for each tissue condition. Shock attenuation was determined for all specimens between the distal radius and proximal ulna accelerometers in the x direction. Missing data from participant 1, 5 and 7 were unusable.

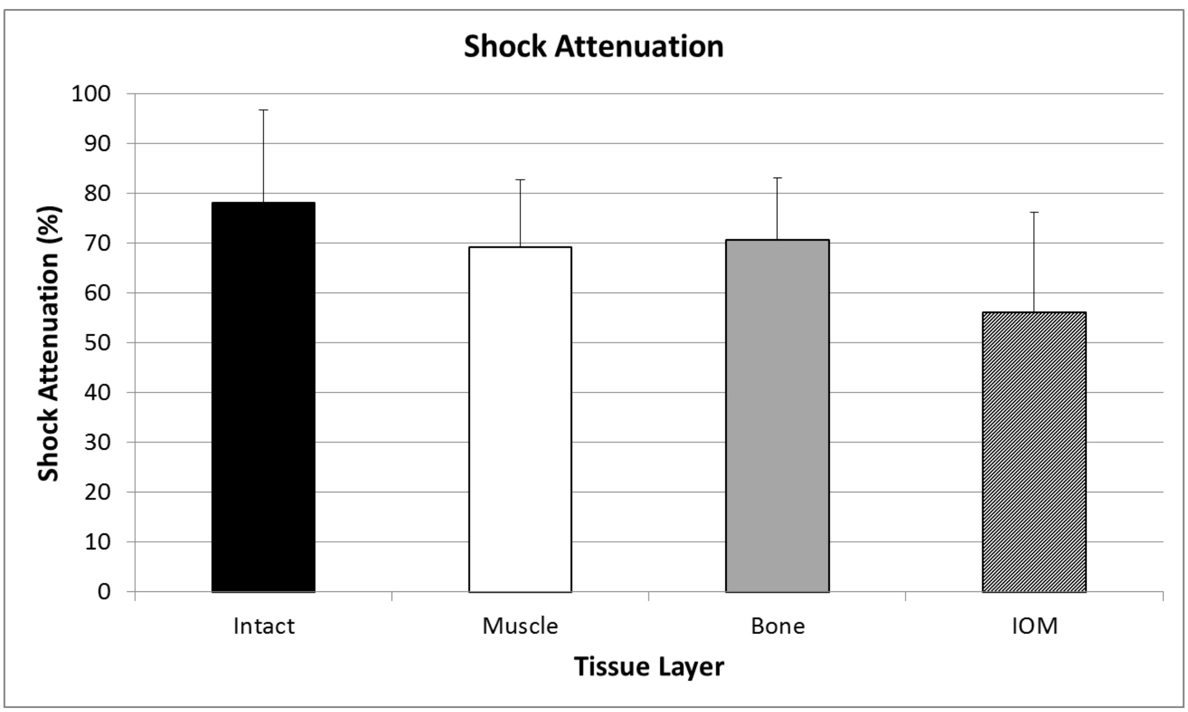

Figure 17: Mean (SD) shock attenuation values for the different tissue conditions.

CHAPTER 5

DISCUSSION

To the author's knowledge, this is the first study to quantify the relative contribution of the skin and adipose tissue, muscle tissue, bone, and interosseous membrane of human cadaver forearm specimens on shock attenuation following a simulated forward fall. Peak acceleration decreased significantly as the shock wave travelled from the distal radius (42.8 (19.0) g) to the proximal ulna (12.3 (4.3) g) in intact human forearm specimens (n=9). Shock attenuation through the forearm decreased non-significantly compared to fully intact specimens when the soft tissues were sequentially removed (skin and adipose: 8.9%; muscle: 7.5% and when the interosseous membrane was cut 21.9%).

5.1 Hypothesis 1

1. The shock wave (acceleration waveform) will be attenuated as it moves from the distal end of the forearm to the proximal end of the forearm. This attenuation will be characterized by a decrease in the magnitude of the peak acceleration and a positive shock attenuation value in the direction parallel to the long axis of the radius and ulna.

Soft tissue oscillations from an impact have been shown to be a significant contributor to the dampening of impact-induced shock waves (reduction in shock wave amplitude as it moves proximally through the body's tissues) (Wakeling & Nigg, 2001). Hypothesis 1 was supported by the overall shock attenuation results (i.e., 78.1% attenuation from distal radius to proximal ulna in the intact tissue condition). Comparable attenuation results have been reported for the lower extremity by Dufek et al. (2009) during running (males 71.7%; females: 83.7%). Although these studies used living participants, the current study used cadaver specimens. Consequently, the attenuation reported herein is the result of the response of passive structures of the body including cartilage, bone, and wobbling mass (Williams et al. 2001).

Overall, the mean peak accelerations decreased in magnitude by a factor of over 3 times as the shock wave travelled from the distal radius (42.8 g) to proximal ulna (12.3 g). Burkhart & Andrews (2010) also found that peak accelerations decreased as the shock wave moved from distal radius (11.1 g) to proximal ulna (5.9 g), while testing the effects of wrist guards for reducing wrist and elbow accelerations. Similarly, Burkhart & Andrews (2013) reported that peak accelerations decreased from the distal forearm (female: 9.5 g; male: 8.9 g) to proximal forearm (female: 6.7 g; male: 6.1 g) when using PULARIS. Although the peak acceleration magnitudes were less in these two studies than those presented in the current study, the same decreasing trends were seen between the distal and proximal forearm. One of the main differences between these studies is that Burkhart and colleagues evaluated living participants whereas non-living specimens were evaluated in this thesis. In living participants, the contraction of the muscles of the forearm would have resulted in an overall stiffer segment than in cadaver specimens, because of their inability to activate muscles. Increased stiffness has been hypothesized previously to reduce shock attenuation in living participants (Butler et al., 2003; Nigg et al., 1995). The greater initial peak accelerations recorded in this study could be caused by higher impact velocities (3.0 - 3.4 m/s) compared to previous work (2.1 - 3.4 m/s in Burkhart, Dunning & Andrews, 2012). Small cracks or fractures in the bone that could not be observed visually after each impact might also help to explain these results. In a study using cadaver radii, Burkhart, Dunning & Andrews (2012) reported mean peak accelerations for pre-fracture, crack, and fracture events that were approximately 40 g, 60 g, and 80 g, respectively. Peak accelerations in the current study were comparable in magnitude, indicating that there could have been some damage to the specimens that could not be seen visually. Fractures were clearly evident in the three female specimens which failed during the initial testing (and were not tested further).

5.2 Hypothesis 2

2. The attenuation between the distal end of the radius and the proximal end of the ulna will be greater than the attenuation between the distal and proximal ends of the radius when the interosseous membrane is intact. When the interosseous membrane is cut, the attenuation between the distal end of the radius and proximal end of the ulna will increase even more.

Since no accelerometer data were available from the proximal radius, a comparison between the proximal and distal radius could not be made. Therefore, the first part of hypothesis 2 could not be addressed.

Cutting the interosseous membrane resulted in a non-significant mean decrease in shock attenuation of 21.9% relative to the intact condition and 14.4% relative to the bone condition. The decrease in mean shock attenuation after the IOM was cut supports hypothesis 2, however it is contrary to previous impact research by Birkbeck et al. (1997). In their work, when the IOM was cut, the proximal and distal load values in each bone equalized, suggesting that, without the IOM, there was no load transfer between the radius and ulna. These contrary findings could be a result of differences in the methods used between the studies. Birbeck et al. (1997) changed the orientation of the distal radio-ulnar joint from complete pronation to complete supination as they increased the load across the joint. The specimens in the current study were fully pronated to simulate how living participants would respond to a forward fall.

5.3 Hypothesis 3

3. As the layers of soft tissue are removed from the forearm, shock attenuation will decrease. Given the relative magnitude of the muscle mass, compared to the other soft tissues, the muscle tissue will have the greatest relative effect on shock attenuation.

Hypothesis 3 was supported because SA decreased on average (by a total of 8.2%) when the soft tissues were removed, compared to the fully intact condition. Muscle tissue did not have the greatest relative effect on SA (a mean decrease of 7.5%) compared to the other tissue conditions (mean SA reduced by 8.9% when the skin and adipose were removed). Cutting the IOM had the greatest relative effect on SA with a mean decrease of 14.4% compared to the bone condition. Soft tissues have been shown previously in multiple research studies involving living participants to attenuate shock (Liu & Nigg, 2000; Pain & Challis, 2004; Chu et al., 1986; Hamill et al., 1995; Lafortune et al., 1996). The attenuation is caused by the oscillation of the soft tissues over the underlying bones (Wakeling & Nigg, 2001). However, each tissue layer has different material properties and would potentially play a different role in shock attenuation (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010). Although this was not statistically supported by the results of this study, there was an overall trend towards decreasing shock attenuation as the tissue layers were sequentially removed. The role that muscle plays in shock attenuation needs to be evaluated further with a larger sample, given the magnitude of the forearm mass attributed to muscle.

Schinkel-Ivy et al. (2012) reported an increase in shock attenuation in participants with a greater wobbling mass to bone mass ratio. They used PA, TPA, and AS to reflect SA in individuals who experienced leg impacts. PA values in this study were between 2 and 6 times lower than those measured in the current study. This is not surprising, when the impact velocities between the two studies are considered. Schinkel-Ivy et al. (2012) impacted participants at between 1.0 and 1.15 m/s, approximately three times lower than the 3.0-3.4 m/s velocities used in this study and in previous work by Burkhart, Dunning, & Andrews (2012) (2.1 m/s – 3.4 m/s).

The malfunctioning accelerometer channels limited the amount of data that could be analyzed, reducing the overall contribution of this work. However, the peak accelerations from the distal radius and proximal ulna accelerometers along the longitudinal axis of the specimens were unaffected by the data collection issues, enabling an evaluation of shock attenuation in this direction. This was the primary focus of this study initially, so despite the limitations presented, it is felt that a positive contribution was made which will drive future work in this area.

Although the current study showed that the contributions of the forearm soft tissues to shock attenuation in in-vitro specimens during simulated forward falls were comparable in magnitude to shock attenuation results in living humans (e.g., for the leg), in-vitro experiments using cadaver specimens are limited in several ways. As previously discussed, cadaver specimens do not have muscle tone, preventing any active shock attenuation to be evaluated. Without muscle tone, the effect that active stiffness of the forearm specimens has on shock attenuation could not be quantified. In addition, the method of impact force application is inconsistent with how impact loads are applied to living humans when they fall. Being able to control the impact conditions between specimens was more important in this initial investigation than trying to more closely mimic the impact conditions of living humans. Despite the care taken, the variability in the magnitudes of the impact forces (not shown) was considerable and needs to be addressed in future work. This variability could have been caused by the specimens not being in the exact same position for each impact even though the impact velocity was kept relatively consistent between trials. Improving the reliability of the impact forces will likely affect the absolute magnitudes and variability of the accelerations collected using this system and protocol. However, the shock attenuation values quantified in the current investigation are a relative measure (distal vs. proximal accelerations) (Dufek et al., 2009), so any differences in impact forces between trials should not affect shock attenuation values differentially.

5.4 Future Directions

Despite the encouraging findings presented regarding shock attenuation, this study should be replicated in the future, once the issues with the accelerometers have been rectified and additional funding has been secured to purchase more specimens. This will help to confirm the results reported here and enable a more thorough evaluation of shock attenuation in the forearm by comparing radius and ulna accelerations in all directions. It is recommended that a jig be developed which could be used to improve the reliability of specimen alignment within the impact system and increase confidence in the impact forces collected. Analyzing the impact forces could also add considerably to the understanding of shock attenuation by the distal upper extremities. Ensuring consistent alignment between the specimens and impactor would also help to rule out this factor as a potential cause of variability in the acceleration data collected more proximally. In addition, the generalizability of the results would be improved if a larger sample of specimens were evaluated. Although the age of the specimens was consistent with an older adult population – a group who is more negatively affected by the outcomes of forward falls – testing younger specimens might help to reduce the age-related declines in bone mechanical properties that were evident in three of the female specimens tested in the current study.

CHAPTER 6

CONCLUSION

In conclusion, this is the first study to quantify the relative contributions of the soft tissues of the forearm to shock attenuation following simulated forward fall impacts. Shock attenuation decreased when the soft tissues were removed from the forearm, with the skin and adipose tissues contributing more than the muscle tissues. Cutting the interosseous membrane had the greatest contribution to decreasing shock attenuation compared to the removal of the soft tissues. Despite the limitations regarding data collection, the reported findings compare favourably to previous work in living humans for the lower extremity and add positively to the current literature in the area, given that it is the first study of its kind. These results could help advance wobbling mass biomechanical models to study the upper extremity impact response of humans following forward falls. In doing so, our understanding of the injury mechanisms associated with impact-induced upper extremity injuries may be improved and strategies to reduce the number and severity of forward fall injuries may be realized.

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APPENDIX A: ANOVA Table

Table 2: ANOVA table for peak acceleration (PA) and shock attenuation (SA).

Peak Acceleration (PA)				
Effect	**df**	**F**	**p**	**(r^2)**
Accelerometer	1	22.30	0.005	0.820
Layer	3	7.64	0.142	0.981
Accelerometer * Layer	3	6.62	0.077	0.869
Shock Attenuation (SA)				
Layer	3	1.82	0.317	0.650

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