The Effects of Impact Loading on the Equine Metacarpophalangeal Joint

by

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ABSTRACT

THE EFFECTS OF IMPACT LOADING ON THE EQUINE METACARPOPHALANGEAL JOINT

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Purpose: To characterize the effect of impact loading on the metacarpophalangeal (MCP) joint and compare the stresses induced by impact and static loading in the equine third metacarpal (MC3) as a first step to assessing the potential role of impact loading in the context of mechanical injury.

Methods: Three analyses were used to characterize impact loading in the equine MCP joint: 1) an ex vivo analysis of vibration attenuation across the joint at a given hoof angle (strike), 2) ex vivo analysis of the contact areas between MC3 and the first phalanx (P1) and proximal sesamoid (PS) bones and 3) finite element (FE) analysis to compare the stress distributions in the distal MC3 of healthy and diseased (osteoarthritic - OA) MCP equine joints under impact and static loading.

Results: Signal energy reaching MC3 was significant at 6-31% of that at the hoof. A heel-first strike produced the largest peak accelerations and highest frequencies among all strike conditions. Contact area between P1 and MC3 was well-defined and bounded by the sagittal and transverse ridges. The ratio of contact area of P1 to PS was 2.83 (P <.0001). Under FE static loading, the highest average stresses (19.38 MPa) were located in the palmar parasagittal groove in the healthy joint. The highest average stresses under impact (14.1 MPa) were located on the dorsal aspect of the medial condyle in the OA model and were greater than the static loading stresses in both models on the dorsal aspect of MC3.
Conclusions: Large accelerations that occur upon impact are attenuated by the equine limb; however still carry considerable energy within the signal that could be damaging to tissue. Contact on MC3 at impact occurs primarily with P1, contrasting with midstance when both P1 and PS are equally involved. MCP FE modeling indicated that an increase in bone stiffness associated with OA may be adaptive under static loading, however increases the stress magnitude under impact loading. Stresses on the distal end of MC3 are comparable to those found during midstance and should be considered in the context of injury.
Dedication

I would like to dedicate this thesis to my late father, Arthur McCarty. His love, support and encouragement has provided me the opportunity to be where I am today and has taught me that no matter how rough the road may be, to never give up.
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# TABLE OF CONTENTS

Abstract..............................................................................................................ii
Dedication..........................................................................................................iv
Acknowledgements..........................................................................................v
Table of Contents..............................................................................................vi
List of Tables.....................................................................................................ix
List of Figures...................................................................................................ix
List of Abbreviations........................................................................................xi

CHAPTER 1: Background and Significance.......................................................1
  1.1 Background...............................................................................................1
  1.2 Joint Anatomy and Function.................................................................2
  1.3 Biomechanics of Joint Loading/ Tissue Mechanics...............................2
      1.3.1 Mechanical and Biological Responses to Loading of Biological Materials.................................................................3
      1.3.2 Mechanical and Biological Responses of Cartilage to Loading ......4
      1.3.3 Subchondral Bone Response.........................................................6
  1.4 Pathophysiology....................................................................................8
  1.5 Contributing Factors to the Development of Osteoarthritis...................9
      1.5.1 Fractures and Bone Remodeling................................................9
      1.5.2 Cellular Response and Signalling.............................................10
      1.5.3 Biomechanical Forces...............................................................11
  1.6 Biomechanical Assessment....................................................................13
      1.6.1 Force Plates and Force Horseshoes.........................................13
      1.6.2 Measuring Joint Contact - Pressure Sensitive Film (PSF) and sensors 14
      1.6.3 Accelerometers......................................................................15
      1.6.4 Finite Element Analysis (Modeling).......................................16
  1.7 Study Aims and Significance...............................................................17
      1.7.1 Aims......................................................................................17
      1.7.2 Hypotheses............................................................................18
      1.7.3 Significance.......................................................................18
  1.8 Figures and Tables.................................................................................20

CHAPTER 2: Effect of Hoof Orientation and Ballast on Acceleration and Vibration in the Hoof and Distal Forelimb Following Simulated Impacts Ex Vivo.........................24
  2.1 Abbreviations.......................................................................................25
  2.2 Abstract...............................................................................................26
  2.3 Introduction...........................................................................................27
  2.4 Materials and Methods.................................................................28
      2.4.1 Specimens...........................................................................28
      2.4.2 Specimen Preparation.............................................................28
      2.4.3 Testing Apparatus.................................................................29
      2.4.4 Accelerometers.....................................................................29
      2.4.5 Hoof Angle at Contact and Limb Ballast.................................30
      2.4.6 Testing Protocol.....................................................................30
### 2.4 Data Analysis
- 2.4.7 Data Analysis .......................................................... 31
- 2.4.8 Statistical Analysis .................................................... 31

### 2.5 Results
- 2.5.1 Alignment of the 3 Accelerometers ............................. 31
- 2.5.2 Angle of Hoof Strike with Respect to the Impactor ......... 32
- 2.5.3 Impact Magnitude and Timing Axially along the Limb (x-axis) ................................................. 32
- 2.5.4 Frequency and Energy of Vibration along the x-axis ........ 32
- 2.5.5 Results in the Transverse (y) and D-P (z) directions ....... 33
- 2.5.6 Changing the Effective Mass ...................................... 33

### 2.6 Discussion
- 2.6.1 Study limitations ...................................................... 34
- 2.6.2 Peak Acceleration .................................................... 34
- 2.6.3 Time to Peak .......................................................... 35
- 2.6.4 Peak Frequency and Frequency at 95% ....................... 35
- 2.6.5 Energy at 95% ........................................................ 36
- 2.6.6 Peak Acceleration Transmission ................................ 37
- 2.6.7 Effective Mass ....................................................... 37
- 2.6.8 Farrier and Bone Injury Implications .......................... 37

### 2.7 Conclusion ............................................................... 38

### 2.8 Figures and Tables ...................................................... 40

### CHAPTER 3: Effect of Hoof Angle on Joint Contact Area in the Equine Metacarpophalangeal Joint Following Simulated Impact Loading Ex Vivo........ 48

#### 3.1 Abbreviations .......................................................... 49

#### 3.2 Abstract ................................................................. 50

#### 3.3 Introduction ........................................................... 51

#### 3.4 Materials and Methods ............................................. 53
- 3.4.1 Specimens ............................................................ 53
- 3.4.2 Specimen Preparation .............................................. 53
- 3.4.3 Placement of Intra-Articular Pressure Film ................. 54
- 3.4.4 Testing Apparatus and Protocol ................................. 54
- 3.4.5 Data Analysis ........................................................ 55
- 3.4.6 Statistical Analysis ............................................... 57

#### 3.5 Results ................................................................. 57
- 3.5.1 Angle of Hoof Strike with Respect to the Impactor ....... 57
- 3.5.2 Contact Areas of the Proximal Phalanx (P1) and Proximal Sesamoid (PS) Bones ................................. 57
- 3.5.3 Lateral and Medial Condyles Contact Area .............. 58
- 3.5.4 Contact Area by Strike ............................................ 58

#### 3.6 Discussion .............................................................. 59
- 3.6.1 Contact Location .................................................... 59
- 3.6.2 Contact Area ........................................................ 60
- 3.6.3 Impact Loading and Joint Degradation ..................... 61
- 3.6.4 Limitations .......................................................... 62

#### 3.7 Conclusion ............................................................. 62

#### 3.8 Figures and Tables .................................................... 63
CHAPTER 4: A Comparison of Stress Distribution in the Healthy and Diseased Equine Third Metacarpal under Impact and Static Loading Using Finite Element Analysis...70

4.1 Abbreviations...........................................................................................................71
4.2 Abstract...................................................................................................................72
4.3 Introduction.............................................................................................................73
4.4 Methods and Materials............................................................................................75
  4.4.1 Image acquisition and segmentation .................................................................75
  4.4.2 Model Features..................................................................................................75
  4.4.3 Mesh Generation ...............................................................................................77
  4.4.4 Material properties ............................................................................................77
  4.4.5 Loading and Boundary Conditions ..................................................................78
  4.4.6 Convergence .....................................................................................................79
  4.4.7 Validation ..........................................................................................................80
  4.4.8 Data Analysis ....................................................................................................80
4.5 Results.....................................................................................................................81
  4.5.1 Convergence .......................................................................................................81
  4.5.2 Validation ...........................................................................................................82
  4.5.3 Static Loading and Stress Distribution ..............................................................83
  4.5.4 Impact Loading Contact Pressure and Contact Area ......................................83
  4.5.5 Impact Loading Stress Distribution ...................................................................83
  4.5.6 Static vs. Impact Loading Contact Pressure ......................................................84
  4.5.7 Static vs. Impact Loading Stress Distribution ....................................................84
4.6 Discussion...............................................................................................................85
  4.6.1 Limitations ........................................................................................................85
  4.6.2 Static Loading and Stress Distribution ..............................................................86
  4.6.3 Contact Pressure and Contact Area ..................................................................87
  4.6.4 Impact Loading and Stress Distribution ............................................................89
  4.6.5 Static vs. Impact Loading Stress Distribution ....................................................89
4.7 Conclusion.............................................................................................................91
4.8 Figures and Tables.................................................................................................92
4.9 Appendices..........................................................................................................102

SUMMARY AND FUTURE DIRECTIONS .......................................................................112

REFERENCES ..............................................................................................................115
LIST OF TABLES

Chapter 1: Background and Significance

1.1 Definitions......................................................................................................................................23

Chapter 2: Effect of Hoof Angle at Primary Impact on Vibration Attenuation in the Equine Distal Forelimb Ex Vivo

2.1 Means and CoV of trials in all locations and directions without ballast........46
2.2 Medians and CI comparing ballast and no ballast for a flat strike.................47

Chapter 3: Effect of Hoof Angle on Joint Contact Area in the Equine Metacarpophalangeal Joint Following Simulated Impact Loading Ex Vivo

3.1 Median and lower/upper limits of the over-limit joint contact area.................69
3.2 Distances between the centroids and the sagittal and transverse ridges ..........69

Chapter 4: A Comparison of Stress Distribution in the Equine Third Metacarpal under Impact and Static Loading Using Finite Element Analysis

4.1 Mesh details used for FE analysis in healthy and OA under static and impact loading.................................................................101
4.2 Number of nodes and elements used in boundary conditions and loading......101
4.3 Mesh details for 3 mesh densities used to determine convergence...............101
4.4 Convergence results based on mesh density by region..............................101

LIST OF FIGURES

Chapter 1: Background and Significance

1.8.1 Metacarpophalangeal (MCP) structure..............................................................20
1.8.2 Factors involved in Equine Osteoarthritis (OA)..............................................21
1.8.3 Phases of the Stance............................................................................................21
1.8.4 Metacarpophalangeal Joint Orientation............................................................22

Chapter 2: Effect of Hoof Angle at Primary Impact on Vibration Attenuation in the Equine Distal Forelimb Ex Vivo

2.8.1 Experimental apparatus (impact hammer set-up).........................................40
2.8.2 Results for peak acceleration (pkacc_x)............................................................41
2.8.3 Results for time to peak (ttp_x)........................................................................42
2.8.4 Results for peak frequency (pkfreq_x)..............................................................43
Chapter 3: Effect of Hoof Angle on Joint Contact Area in the Equine Metacarpophalangeal Joint Following Simulated Impact Loading Ex Vivo

3.8.1 Experimental apparatus (impact hammer set-up).................................63
3.8.2 Placement of pressure sensitive film within the MCP joint.........................64
3.8.3 Sample film (thresholded, calibrated and segmented into pressure levels).....65
3.8.4 Global mean and range depicted on sample films (medial and lateral)........66
3.8.5 Comparison of midstance and impact contact areas on P1 and MC3..........67
3.8.6 Comparison of P1 contact area according to hoof strike.........................68

Chapter 4: A Comparison of Stress Distribution in the Equine Third Metacarpal under Impact and Static Loading Using Finite Element Analysis

4.8.1 Phases of the stance.................................................................................92
4.8.2 Area defined by location of the distal end of MC3 used for the data analysis...92
4.8.3 Mapped material stiffness within FE models based on CT image bone density..93
4.8.4 Material mapping at multiple mesh densities testing to determine convergence..94
4.8.5 Comparison of multiple density-modulus equations used to map material properties..........................................................95
4.8.6a Comparison of stress between static loading in current study and previous study.............................................................................95
4.8.6b Comparison of average stress under static loading by location at 30° palmar...96
4.8.7a Comparison of impact contact pressures in both FE models and experimental data............................................................................96
4.8.7b Comparison of average impact contact pressures by location..................97
4.8.8a Comparison of surface stress between healthy and OA models under impact...97
4.8.8b Comparison of average impact stress by location ......................................98
4.8.9 Comparison of average contact pressure between current impact data and previously reported static data.........................................................98
4.8.10a Comparison of surface stress under impact and static loading ..............99
4.8.10b Comparison of average stress under impact and static loading by location in a palmar slice...........................................................................99
4.8.10c Comparison of average stress under impact and static loading by location in a dorsal slice.................................................................100
4.9 Appendices (A/B) ..................................................................................102
LIST OF ABBREVIATIONS

MCP – Metacarpophalangeal Joint

OA – Osteoarthritis

MC3 – Third metacarpal bone

P1 – First phalanx bone

PS – Proximal sesamoids

SCB – Subchondral bone

1° – Primary

2° – Secondary

GRF – Ground reaction force

PSF – Pressure sensitive film

FE – Finite element

FEA – Finite element analysis

FEM – Finite element modeling

CT – Computed tomography
CHAPTER 1: BACKGROUND AND SIGNIFICANCE

The aim of this research is to compare the stresses induced in the equine third metacarpal (MC3) by impact and static loading, as a first step for comparing the relative roles of both modes of loading in the etiology of injury and joint disease. To achieve this aim, a broader analysis is necessary of impact on the metacarpophalangeal (MCP) joint, including the attenuation of vibration across the joint, and the areas of contact with the first phalanx (P1) and proximal sesamoid (PS) bones.

1.1 Background

Horse racing is one of the most physiologically demanding sports within the equine industry and most racehorses experience some degree of lameness and injury during their careers. The most common site of injury in the forelimb of racehorses is the MCP joint (Sandgren et al. 1993, Strand et al. 1998, Easton et al. 2007). This joint is particularly susceptible to injury because of the large range of motion in the joint, the relatively small surface area with respect to body size and the magnitude of mechanical loading it experiences during high-speed movement (McIlwraith 1996). Approximately 25% of racing Thoroughbreds in North America experience pain associated with lameness in the MCP joint, which is the leading cause of financial loss in the racing industry (Cruz et al. 2008). Injuries range from mild synovitis to severe osteoarthritis (OA), and can be associated with catastrophic condylar failure (fracture through the joint surface due to a single high-overload event). Some injuries are due to unavoidable impact however most are related to exercise-induced cyclical trauma due to the inability of the musculoskeletal system to adapt rapidly to the training load (Firth 2008). In
Ontario, 1/3 of all 2 and 3 year old horses in a study of 50 horses had OA in the MCP joint and the severity of OA increased with age, up to the age of 6 (Neundorf et al. 2010).

1.2 Joint Anatomy and Function

The MCP joint is formed between MC3, P1 and the PS bones (Figure 1.8.1). It is evolved to produce a smooth, essentially frictionless motion during movement and to effectively transfer loads between P1, PS and MC3. As for all synovial joints, the articular surface of each bone is covered with a thin layer of hyaline cartilage and is contained within a joint capsule that is lubricated by synovial fluid. Beneath the articular cartilage is a layer of calcified cartilage, which provides a transition between the stiff subchondral bone (SCB) and the compliant articular cartilage. SCB can be further classified into the subchondral bone plate and underlying trabecular bone. MC3 and P1 are connected by medial and lateral collateral ligaments, while additional collateral ligaments attach the PS to the sides of the MC3 condyles and proximal end of P1. Sesamoidean ligaments connect the bases of PS to P1 to ensure that the 3 bones move in unison against MC3. The sesamoid bones have extensive ligamentous support to aid in supporting the joint as the sesamoids slip to the distal aspect of the MC3 condyles in maximal extension (dorsiflexion) of the joint that occurs during high-speed movement. Large dorsal and caudal pouches of the MCP joint capsule that lie against the MC3 shaft allow for the high degree of flexibility and range of motion (Dyce 2010).

1.3 Biomechanics of Joint Loading/ Tissue Mechanics

A normal functioning MCP joint requires the proper action of the synovial fluid, articular cartilage, subchondral bone and has a very low coefficient of friction (experimentally determined to be ~0.007; Noble et al. 2011). The individual action of the structures involved in the joint capsule allows the joint to move fluidly while absorbing and dissipating the applied load during
extreme weight bearing phases of the stride. Normal loads experienced under light exercise are known to stimulate the synthesis of proteoglycan and selectively increase the compressibility of the tissues involved, effectively increasing the shock absorbing capabilities (McIlwraith 1996). Overloading the joint (as can occur during repeated, extreme extension of the MCP in horse galloping at full speed for an extended period of time) can have the opposite effect and may eventually lead to structural breakdown.

1.3.1 Mechanical and Biological Responses to Loading of Biological Materials

The effect of mechanical loading on biological material has both an instantaneous physical response as well as biological (or adaptive) response that occurs within the tissue over time. The instantaneous response to loading is largely determined by the material properties of a tissue including: structure (isotropic vs. anisotropic), material response (elastic, viscoelastic, plastic), type of material (poroelastic, biphasic), measurable material response (stress, yield stress, strain) and other known properties that determine how a material responds when loaded (Young’s Modulus, Poisson’s Ratio, Aggregate Modulus) (Table 1.1). These properties dictate how the tissue will respond instantaneously when mechanically loaded and respond accordingly depending on the loading magnitude and loading rate.

The biological response to loading over time elicits a cascade of events that is responsible for maintaining tissue health either through growth and/or repair. In bone tissue, the biological response to loading over time refers to the geometry and microarchitecture of a bone and how bone modelling and bone remodelling are involved in the growth and repair as a response to the mechanical loading that occurs over time. Bone modeling is defined as bone formation or resorption at a given site to produce functionally and mechanically adapted architecture,
compared to *remodeling* where small areas of damaged bone are removed by osteoclasts and replaced by osteoblasts, essentially removing the defective bone matrix and replacing it with viable bone tissue (Frost 1994).

Mechanical properties of the stiffness and strength of the entire bone are dependent on bone quality (properties of the bone material), bone shape and the direction of loading. Bone porosity is highly related to the mechanical properties of bone tissue however, bone architecture/structure is also a determining factor that affects the biomechanical response to loading (Rubio-Martinez et al. 2010). Bone porosity varies greatly between bone types with less than 15% porosity in cortical bone compared to the greater than 70% porosity found in trabecular bone (Schaffler et al. 1988). Bone strength and stiffness vary inversely with increasing porosity (Carter et al. 1977). Trabecular bone microarchitecture refers to structure based on the shape, width, connectivity and anisotropy within trabecular bone and is a main structural determinant of mechanical strength in bone (Dalle Carbonare et al. 2004). High-strains recorded on the MC3 in exercised horses were found to be more related to the shape of the bone rather than the parameters of the bone quality (Davies 2001). There is some debate as to which property (bone shape or quality) has a greater effect on the mechanical properties of bone however these properties cannot act alone in determining the overall strength of a whole bone. Overall, bone function relies on the material properties that make up bone and how it is put together (structure), where structure determines shape and size of the bone (Les 1997).

1.3.2 Mechanical and Biological Responses of Cartilage to Loading

The cartilage matrix is a poroelastic, biphasic material (Table 1.1), made up of the collagen-proteoglycan component and an interstitial fluid component. These properties are defined by the permeability, aggregate modulus and Poisson’s ratio (Table 1.1) (McIlwraith
Under normal joint loading, the cartilage deforms to maximize the contact area thereby reducing the amount of stress within the cartilage (Brandt et al. 2009). When the joint is loaded, the fluid component of the cartilage is acted upon first and then the force is fully transferred to the solid component of the cartilage once a new equilibrium point has been reached. Joint loading will cause instantaneous cartilage deformation followed by a slower creep (Table 1.1) as fluid moves out of the matrix until the material is fully compressed. The degree of deformation is dependent on the magnitude, rate and the duration of the load applied. The biological response of chondrocytes embedded in the cartilage will differ depending on the loading conditions such as those that occur in quasi-static loading (i.e. time dependent but enough such that inertial effects can be ignored) or impact loading (high force or acceleration loading that occurs over a short period of time). Healthy chondrocytes respond to mechanical loading by maintaining the biophysical properties of the extracellular matrix. Quasi-static loading will generally inhibit biosynthetic activity, while moderate cyclic loading will increase biosynthetic activity in order to adapt to the loading (McIlwraith 1996). High-frequency, low-magnitude loading has been shown to improve chondrocyte proliferation, however frequencies above or below the optimal level have been shown to inhibit DNA synthesis (Liu et al. 2001). Even when loaded within the optimal frequency range, an increase in the loading magnitude tended to be associated with inhibition of the proliferation response and led to chondrocyte death (Milentijevic et al. 2003). High dynamic loading (impact) is thought to dramatically inhibit protein and proteoglycan synthesis allowing for degradation without repair (Radin et al. 1984, Jeffrey et al. 1995). Although high-magnitude stresses > 20 MPa have been shown to cause chondrocyte cell death (Green et al. 2005, Loening et al. 2000), this response was only found to occur in conjunction with high loading rates as no evident damage was found when the same peak stresses were
attained at a slower strain rate (Quinn et al. 2001). It has been shown that chondrocytes that survive cartilage impact loading as a result of a decrease in loading magnitude may still contribute to cartilage matrix degeneration through the secretion of degradative factors and chondrocyte cell deactivation (Pickvance et al. 1993, Quinn et al. 2001) and therefore it has been suggested that the rate of loading may be more injurious to cartilage than the magnitude of loading applied (Brandt et al. 2009). Under physiological high-rate loading, the articular cartilage stiffens (to a degree dependant on the loading rate), decreasing the strain (Table 1.1) and strain rate (compared to strain at quasi-static loading), thereby protecting chondrocytes from injury (Langelier et al. 2003). Under very high-loading rates (like those experienced in the MCP joint of a horse moving at racing speeds), the elastic (energy storage) response increases with loading frequency, while the viscous (energy dissipation) response remains the same (Fulcher et al. 2009). This means that at higher frequencies, more energy is stored in the cartilage than is dissipated and will likely lead to the formation of cracks within the cartilage as a method of releasing the stored energy. Measurement of MCP joint pressures indicates that, as the loading magnitude and rate increases with speed, contact areas also increase, suggesting that areas not normally loaded may not be as well conditioned to handle the extreme loading at high-speed movement which could cause damage to the articular cartilage (Easton et al. 2007).

1.3.3 Subchondral Bone Response

SCB is a poroelastic and viscoelastic (Table 1.1) material that has a time-dependent response to an applied load. SCB can range in porosity from 50-90% (compared to cortical bone which is typically less than 30% porous) and is able to deform to compressive strain by over 50% (McIlwraith 1996). A large strain is indicative of a bones’ ability to absorb a considerable amount of energy under a large compressive load. SCB adapts to stresses endured during joint
loading by increasing bone density and strength through bone modeling and remodeling. Bone modeling and bone remodeling is determined by the magnitude, area of stress concentration, rate and duration of an applied load and varies with the type of loading. Stress within the MCP been shown to change as contact areas changes under increasing loads (Brama et al. 2001). Contact area increases under higher loading and suggests that areas not normally loaded in low-load conditions (static loading during walking/standing), are not conditioned to regular loading, are then subjected to a high degree of stress during high-force loading such as galloping (Easton et al. 2007). The effect of loading rate on the mechanical response of bone, including yield strength (or yield stress), is based on the viscoelastic nature of SCB material, specifically the collagen within the bone matrix. The Young’s Modulus (Table 1.1) of SCB is lower at low loading rates than high loading rates because the collagen within the bone exhibits higher stiffness and is more brittle under high loading rates, therefore decreasing the fracture toughness and the ability to resist structural failure (Kulin et al. 2008).

The role of SCB within a joint is to help distribute axial loads associated with movement in order to spare the overlying articular cartilage from damage. This is achieved through the orientation of SCB trabeculae, which are organized so as to provide maximum strength to the articular cartilage while still maintaining energy absorbing properties. Increased trabecular compliance allows a greater degree of strain (Table 1.1) when loads are applied to the joint, effectively increasing the energy absorbing capabilities. Although SCB is almost ten times stiffer than cartilage, it is much softer than cortical bone and serves as the major shock absorber because the cartilage is too thin to act as an effective shock absorber (Brandt et al. 2009). In addition to the SCB, there is a calcified cartilage layer which acts as a transitional zone between
the much stiffer SCB and the compliant articular cartilage whose intermediate stiffness is responsible for transforming shear stresses into compressive and tensile stresses which the articular cartilage is better able to withstand (Redler et al. 1975). Under impact loading, the degree of SBC deformation is restricted due to its viscoelastic nature. Failure to properly absorb impact leads to an accumulation of microdamage within the SCB plate and calcified cartilage, initiating the remodeling response which in turn leads to trabecular sclerosis or thickening (Frost et al. 1983, Brandt et al. 2009). Sclerosis increases the stiffness of the bone and limits the shock absorbing capabilities of the SCB thereby reducing the chondroprotective effect (Norrdin et al. 1998, Brandt et al. 2009).

1.4 Pathophysiology

Traumatic arthritis (osteoarthritis) is failed repair of damage through a pathological response to excessive mechanical stress caused by biomechanical overload that typically occurs in high motion joints (Radin 1972, Brandt et al. 2009). Osteoarthritis (OA) is a progressive and permanent deterioration of the articular cartilage within a joint that is accompanied by changes in the bone and soft tissues (McIlwraith 1996) (Figure 1.8.2). OA is typically characterized by local fibrillation or complete erosion of the articular cartilage, sclerosis of the subchondral bone plate and trabecular bone, chronic synovitis (inflammation), osteophytes, pain and reduced joint movement due to the thickening of the joint capsule. Joint disruption can occur by either affecting the articular cartilage of the joint or the underlying SCB. Fatigue or damage to the collagen framework of cartilage, can expose chondrocytes to physical trauma that causes further injury and metabolic change. Accumulated damage to SCB and calcified cartilage causes a reactivation of the secondary center of ossification, advancing the tidemark towards the articular surface through endochondral ossification, thereby leading to thickening of the mineralized
tissues and thinning of the overlying articular cartilage (Burr and Radin 2003). In addition to the structural changes, the microdamage occurring in the SCB action can trigger a metabolic response which causes a release of cytokines that in turn damages the articular cartilage through enzymatic degradation (McIlwraith 1996). Many studies have found that when there is significant degeneration of the articular cartilage, it is accompanied by sclerotic SCB but there are at least two mechanisms; a bone injury pathway and an articular cartilage pathway that lead to changes in SCB and OA (Cruz et al. 2008).

1.5 Contributing Factors to the Development of OA

Development of OA in the MCP joint of horses can occur in a previously normal joint after injury, or where there is a joint conformational abnormality (Figure 1.8.2). Both occurrences predispose the joint to disruption caused by one or a combination of the following: 1) physical forces associated with the biomechanics of high-speed movement and the biomaterial failure of the articular cartilage and subchondral bone, 2) failure of chondrocyte cellular signalling and response to degradation and repair, 3) bony remodeling, microfractures, and changes to the vascularization of the joint tissues (McIlwraith 1996).

1.5.1 Fractures and Bone Remodeling

Bone responds to mechanical stimuli by modeling and remodeling to strengthen the areas in the direction that the primary load is applied (Frost et al. 1990, Easton et al. 2007). Remodeling is also responsible for controlling bone shape and size, but morphological remodeling differs from the remodeling involved in the manipulating the micro-architecture of the bone and controlling bone quality (Martin and Burr 1989). Approximately 10-30% of remodeling targets micro-damage repair (Burr 2002) and is initiated by microcracks in the SCB
and calcified cartilage accounting for the increase in vascularization within OA joints (Brandt et al. 2009). Horses with OA develop a persistent resorptive response where bone formation and resorption are uncoupled, resulting in remodeling that is sustained for months and can lead to irreversible and permanent loss of bone architecture (Cruz et al. 2008). This dysregulation of bone metabolism leads to focal radiolucent areas that can undermine and penetrate the SCB plate resulting in a loss of SCB support and invasion into the calcified cartilage (Cruz et al. 2008), weakening the internal structure of the bone through crack propagation by increasing porosity in the area (Muir et al. 2006). In equine OA, the distal end of MC3 is characterized by areas of high bone density adjacent to areas of increased porosity and are thought to create high stress gradients within the joint (Rubio-Martinez et al. 2008). Joint-pressure measurements have verified that areas consistently in contact under higher load are associated with increased subchondral bone density suggesting that SCB remodels and adapts to the applied load (Easton et al. 2007). Although stiffening of the SCB occurs in OA, previous data suggest that based on these changes alone, the increase in stress due to the increase in bone stiffness would not likely account for the destruction of the articular cartilage (Burr and Radin 2003, Brown et al. 1984).

1.5.2 Cellular Response and Signalling

Signaling from osteocytes undergoing apoptosis adjacent to sites of micro-damage may form part of the targeting mechanism for remodeling and repair (Verborgt et al. 2002; Noble et al. 2003). The presence of inflammation within the joint due to traumatic synovitis has deleterious effects on the joint capsule. Chronic inflammation and repeated injury can lead to vascular damage that contributes to SCB necrosis. The repeated overextension and impact of P1 on MC3 causes generalized capsulitis and synovitis as well as impingement of bones that result in osteophytes and direct cartilage injury. Chemokines in the synovial fluid and synovial
membrane drive catabolic metabolism in the cartilage and contribute to the loss of biochemical constituents of cartilage as well as collagen network damage. Healthy chondrocytes become entrapped in the collagenous matrix where the damage occurs and are unable to help with repair and eventually die (McIlwraith 1996). A combination of cellular apoptosis, loss of cell populations and a lack of cellular repair, contributes to the loss of biomechanical properties within the MCP joint. This process leads to irreversible cartilaginous lesions that continue to expand if the joint is continually subjected to large magnitudes and high frequencies.

1.5.3 Biomechanical Forces

Training and racing can lead to mild yet chronic biomechanical overloads and MCP joint injury. The mechanical interactions between the horse’s hoof and the ground can best be described through the phases of the stance (Figure 1.8.3) where each phase demonstrates different biomechanical characteristics. The first phase of the stance is impact, which can be divided into primary (1°) and secondary (2°) impact events. Primary impact begins when the hoof collides with the ground and accounts for approximately the first 7% of the stance (Thomason et al. 2008). This phase is characterized by large decelerations and high-frequency vibrations that are transmitted from the hoof through the bony structures of the distal limb (Gustás et al. 2006, Drevemo et al. 1994). The combination of high-frequency vibrations and rapidly increasing ground reaction forces (GRFs) is thought to be partially responsible for accumulated microcracks to SCB (Radin et al. 1973). However, laboratory testing on equine cadaver limbs reveals that the magnitude of acceleration signals on MC3 are reduced to approximately 10% of that at the hoof suggesting there is significant vibration attenuation across the interphalangeal and MCP joint (Lanovaz et al. 1998). There is a large vertical deceleration with some horizontal deceleration and minimal GRFs during primary impact. The forces are
determined by the rate of deceleration and mass involved (during primary impact this includes
the mass of the hoof and a small portion of the distal limb). Secondary impact accounts for
approximately the next 5-30% of the stance and is characterized by the first stage of the collision
of the mass of the horse with the leg as it becomes firmly planted on the ground (Thomason et al.
2008). During this phase there is a significant increase in force and strain due to a large increase
in horizontal acceleration and rapidly increasing GRFs, as a result of the forward momentum of
the body mass onto the limb and the bones of the limb pushing forward onto the hoof. The third
stage of the stance, midstance, accounts for approximately 5-90% of the stance following impact
depending on the speed and gait (Thomason et al. 2008). During midstance, the full body weight
of the horse is smoothly loaded on the limb and while vertical GRFs are at their peak,
accelerations (both vertical and horizontal) are no longer acting on the system. The extremely
high vertical GRF during midstance, upwards of 2.5 times the body weight, is likely to contribute
to many bony and soft tissue injuries (Witte et al.2004). Injury occurs under the increased
loading not only because of the extremely high vertical GRF, but also because of the
biomechanical action of the MCP joint during high-speed movement. The angle of the MCP
joint at midstance can be close to 90° under high-speed movement and contributes to the
direction of the loading at this phase of the stance (Figure 1.8.4). Joint orientation is an
important consideration to keep in mind when evaluating and comparing loading mechanics
(impact vs. quasi-static loading) within a joint in determining the contact location and area
(Figure 1.8.4). The peak vertical force is parallel to MC3 and primarily exerts a compressive
stress on the bone. As speed increases and the horse begins to fatigue, there is increased
dorsiflexion in MCP, which when the horse is galloping has been shown to increase the bending
stress on MC3 (Les et al 1997). Breakover is the final stage of the stance and is approximately
the final 85-100% of the stance and is the phase where the hoof lifts at the heels and rolls off from the ground (Thomason et al. 2008). Breakover is characterized by rapidly decreasing forces by unloading the limb.

Abnormal joint loading can result from conformational abnormality, lameness in a contralateral limb and musculoskeletal fatigue. In addition, chronic cyclic joint loading can also lead to abnormal bone adaptation, including sclerotic or necrotic bone, and erosion of the articular cartilage, which then predisposes the joint to further injury. Loading magnitude, rate and duration have all been shown to drive bone formation, and each has a direct relationship with the rate of bone remodeling (Rubin and Lanyon 1987; Cruz et al. 2008, Kulin et al. 2008). During mid-stance quasi-static loading, forces from P1 and PS acting on the condyles of MC3 create a nutcracker effect that can lead to SCB failure as a result of the sesamoid bones producing a compression force and a shear force after impact of the hoof with the ground during high speed movement (Norrdin et al.1998). Unlike static loading, dynamic (impact) loading has been shown to cause an increase in bone formation (Rubin and Lanyon 1987). Quantification of how joint contact changes under different types of loading conditions and the adaptation of the bone to this change in normal and abnormal joints may provide further insight into the pathogenesis of OA.

1.6 Biomechanical Assessment

1.6.1 Force Plates and Force Horseshoes

Force plates are instruments used to measure the ground reaction forces generated by a body standing on, or moving across, to quantify balance, gait and other parameters of locomotory biomechanics. While force plates are practical for assessing equine motion at slower
gaits (walk, trot), higher speed motions (racing trot, canter, gallop) are difficult to accurately measure due the small size of the plate and increased stride length making it difficult for the horse to land in the center (Roland et al. 2005). In order to be able to study multiple footfalls over varying speeds, dynamometric horseshoes have been designed and tested to measure force in the moving horse (Roland et al. 2005, Setterbo et al. 2009, Robin et al. 2009). Although the horseshoe is able to provide additional data, these shoes are typically heavy and can influence the natural gait pattern.

1.6.2 Measuring Joint Contact - Pressure Sensitive Film (PSF) and sensors

Current methods that have been used to study equine joint contact include the dye staining technique, use of pressure sensitive film (PSF) and pressure sensors. The method chosen is largely dependent on the type of joint loading being applied and the level of joint disruption tolerable. While not all methods are invasive, those that do require a level of joint disruption are limited to use in an ex vivo environment.

Dye staining technique has been used to study the equine MCP joint contact area (Brama et al. 2001, Easton et al. 2007) and does not require disruption of the joint. While this method can be used in an in vivo environment, it does not provide absolute pressure values and is unable to distinguish between the degrees of loading (Easton et al. 2007).

PSF is a highly sensitive film that reacts according to the level of pressure applied which is indicated by the density of the color change showing the pressure distribution and magnitude between any two contacting, mating or impacting surfaces. PSF have been used previously in equine research to determine the contact area of a joint surface under varying loading conditions and to determine changes in joint contact that occurs with joint disease
(Colahan et al. 1987, Brama et al. 2001, Bowker et al. 2001, Hartog et al. 2009). There is some inherent error associated with PSF due to the required joint disruption in order to insert the film into the joint and is therefore only appropriate to use ex vivo testing.

Pressure sensors are used in joints, to determine loading force, pressure, and contact area, and can provide real-time data. Pressure sensors have been used previously to evaluate pressure mapping within the equine MCP joint (Easton 2012). However, due to the highly curved configuration of MC3, many of the sensors become bent within the joint leading to inaccurate data recordings. In addition, the high forces generated under midstance loading, were found to overload the sensors and no data could be collected from that particular area (Easton 2012). While pressure sensors have been found to be effective under quasi-static loading, high rate loading has been found to cause the sensors to fail, even if the load magnitudes are within range (Dr. K. Easton, personal communication).

1.6.3 Accelerometers

A triaxial accelerometer directly measures acceleration along 3 orthogonal axes (x, y, and z), where the orientation of the accelerometer on a specimen determines the direction of the accelerations measured. Accelerometers are commercially available and ranges of response (+/-5000m/s²) and frequency (0-5000 Hz) suitable for locomotory experiments. When paired with data acquisition systems that are able to sample at high rates, they are appropriate for recording acceleration during high speed motion (racing trot and gallop). They have also been used to characterize time dependent events such as impact, as well as the subsequent vibration of biological tissues including hoof and bone (Thomason et al. 2008, Gustås et al. 2006, Lanovaz et al. 1998, Chateau et al. 2010). Accelerometers can be used to examine many factors associated

1.6.4 Finite Element Analysis (Modeling)

Finite element modeling (FEM) or analysis (FEA) is a numerical method for solving problems of engineering and mathematical physics. It is generally not possible to obtain an analytical solution (one requiring the solution of ordinary or partial differential equations) because many of these problems involve complex geometries, loadings, and material properties. In situations such as these, a numerical method such as FEM is used. FEM results in a system of simultaneous algebraic equations for a solution that yield approximate values of the unknowns at discrete points (also known as nodes or nodal points) on a continuum (Logan 1986). Instead of solving the problem for the entire object in one operation, the object is divided into equivalent systems of smaller units, the finite elements (FE). FEM calculates displacements, stresses and strains for each finite element and combines the elements to obtain a solution for the entire object. The use of FEM in biological applications can be a non-destructive method of predicting stresses and strains in structures that have complex geometry, specific material properties and undergo complex loading.

In biological tissues, FEM is often based on data collected via computed tomography (CT) and/or magnetic resonance imaging (MRI). These diagnostic images are used to create a virtual model that is subject specific in both material properties (i.e. bone density) and geometry (i.e. micro-architecture). By assigning the material properties, such as Young’s modulus and
Poisson’s ratio, each model behaves according to the user-defined parameters under a specified loading condition. FE analysis of these computer simulated models allows for the comparison of multiple loading conditions in order to determine stresses and strains unique to a subject without performing destructive mechanical testing or invasive live animal testing, and allows for repeated trials while incorporating a number of variables that can be changed to influence the study outcomes.

The FE method has been used within equine research to examine the stress distribution in the hoof (Hinterhofer 1997/2001, McClinchey 2003) proximal phalanx (O’Hara 2012), third metacarpal (Les 1997, Easton 2012) and entire distal forelimb (Collins 2009). While these studies have provided valuable data including stress distribution within bone (O’Hara 2012, Les 1997, Easton 2012), tendon (Collins 2009), hoof (Hinterhofer 1997/2001, McClinchey 2003, Collins 2009) and the effect of bone geometry on contact stress within the MCP joint under quasi-static loading, to the author’s knowledge, there is no FE equine model that examines the effect of impact loading. Although FE models are not currently used as a diagnostic tool among practitioners to assess bone and joint health, it is possible that this method of biomechanical analysis could provide insight into the effect of biomechanical impact loading within a joint and can provide predictions for in vivo parameters that could not be measured otherwise.

1.7 Study Purpose and Significance

1.7.1 Aim

The overall aims of these experiments were to:

1. Characterize bone accelerations of MC3 under impact loading ex vivo.
2. Determine the contact area and contact pressures in the MCP joint under impact loading ex vivo.

3. Compare the stress distribution in the distal end of MC3 under impact and static loading in healthy and diseased (OA) MCP equine joints using FE analysis.

1.7.2 Hypothesis

The biomechanical loading that occurs during impact creates stresses comparable to those during midstance in the subchondral bone of the distal aspect of the third metacarpal in the equine metacarpophalangeal joint.

1.7.3 Significance

Current Ontario death records indicate that there is a significant amount of horse wastage due to injury. Data on 137 Standardbreds in the Ontario death registry program in 2004-5 showed that 37 (27%) were destroyed because of musculoskeletal injury, 18 of them due to fracture of the first phalanx, P1 (unpublished data courtesy of Dr. Antonio Cruz). Fractures to the condyles of MC3 and the proximal sesamoids in the forelimb are the top two fracture site locations among Thoroughbred racehorses in the United Kingdom and the United States of America (Parkin 2004, Johnson 1994, Cohen 2000). Many factors contribute to the cause of these injuries that lead to the increased fatalities seen on the racetracks including track surface and design, and the biomechanical stress placed on the body. Previous studies (Radin et al. 1973; Serink et al. 1977) have indicated that impact shock contributes to SCB damage and extrinsic factors that can be influenced are significant for study as a potential to reduce the risk of injury in the racing industry. The equine industry will benefit from not only improving animal welfare, by reducing the risk of injury and minimizing injury related track deaths, but also financially, by prolonging
individual horses racing careers and reducing the amount of wastage due to injury. Currently, there is a lack of evidence that has evaluated the biomechanics during multiple phases of the stance and their role on MCP joint loading and injury. While static loading has been well studied, the role of impact on joint injury in horses remains relatively unknown. In examining the effect of impact on the biomechanical properties of the MCP joint, it is possible that this research will provide some insight into the role of impact on the development of OA.
1.8 Figures and Tables

**Figure 1.8.1** – The bones that make up the equine distal forelimb including the metacarpophalangeal (MCP) joint (circled in red). A parasagittal (or longitudinal)-sectional slice of the MCP joint indicating the structures present within the joint capsule (images adapted from Thomas 2005).
Figure 1.8.2 – Factors involved in equine osteoarthritis that are associated with changes to the articular cartilage of the metacarpophalangeal joint. Figure reproduced from McIlwraith 1996.

Figure 1.8.3 - Phases of the Stance: Blue vector = Ground Reaction Force (GRF) and red vector = Accelerations (Vertical and Horizontal). Figure reproduced from Thomason and Peterson 2008.
**Figure 1.8.4** – Joint Orientation during (1) metacarpophalangeal joint angle at impact and (2) MCP joint angle at midstance phase of the stride.
<table>
<thead>
<tr>
<th>Mechanical Response</th>
<th>Definition</th>
</tr>
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<tbody>
<tr>
<td>Biphasic</td>
<td>A material that has both a solid phase and a fluid phase.</td>
</tr>
<tr>
<td>Poroelastic</td>
<td>A material that has physical properties which are determined by those of its constituent material and by the volume occupied by pores.</td>
</tr>
<tr>
<td>Elastic</td>
<td>The deformation of a material that exhibits instantaneous and completely reversible deformation once the forces are no longer applied - the object returns to its original shape.</td>
</tr>
<tr>
<td>Viscoelastic</td>
<td>Materials that have some properties of both a viscous fluid and an elastic solid and exhibit time dependent behaviors of creep and stress relaxation when forces are applied and removed.</td>
</tr>
<tr>
<td>Plastic</td>
<td>The deformation of a material that has undergone non-reversible changes of shape in response to applied forces.</td>
</tr>
<tr>
<td>Isotropic</td>
<td>A material that has material properties that are independent of direction and only have 2 independent variables or elastic constants (Young’s modulus and Poisson’s ratio) in their stiffness and compliance matrices.</td>
</tr>
<tr>
<td>Anisotropic</td>
<td>A material that has material properties that are dependent on the direction, having 21 elastic constants in their stiffness and compliance matrices.</td>
</tr>
<tr>
<td>Permeability</td>
<td>Rate of fluid flow through a material (cartilage matrix).</td>
</tr>
<tr>
<td>Strain</td>
<td>A measurement of deformation</td>
</tr>
<tr>
<td>Stress</td>
<td>A measure of force per unit area</td>
</tr>
<tr>
<td>Yield Stress/Strength/Point</td>
<td>The stress at which a material begins to deform plastically. Once the yield point is passed, some fraction of the deformation will be permanent and non-reversible.</td>
</tr>
<tr>
<td>Creep</td>
<td>The increase in strain exhibited by a material under constant loading as the time increases.</td>
</tr>
<tr>
<td>Stiffness</td>
<td>The extent to which a material resists deformation in response to an applied force.</td>
</tr>
<tr>
<td>Aggregate Modulus</td>
<td>The compressive stiffness of the solid matrix (cartilage).</td>
</tr>
<tr>
<td>Poisson’s Ratio</td>
<td>The ratio of transverse contraction strain to longitudinal extension strain in the direction of stretching force</td>
</tr>
<tr>
<td>Young’s Modulus</td>
<td>One of the elastic moduli that is used as a measure of the stiffness of an elastic isotropic material.</td>
</tr>
</tbody>
</table>

**Table 1.1** – Definitions for all mechanical responses associated with tissue mechanics (Cowin 2007).
CHAPTER 2: EFFECT OF HOOF ANGLE AT PRIMARY IMPACT ON VIBRATION ATTENUATION IN THE EQUINE DISTAL FORELIMB EX VIVO.

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**Authors and Contributions:**

Cristin A. McCarty – *The University of Guelph (Department of Biomedical Science)* The main contributor to study design, did the majority of the study execution, the majority of the data analysis and interpretation, and the majority of the manuscript preparation.

Jeffrey J. Thomason - *The University of Guelph (Department of Biomedical Science)* Made a significant contribution to the study design, a small contribution to the study execution, a significant contribution to the data analysis and interpretation (programming in Matlab) and provided guidance on manuscript preparation.

Karen Gordon - *The University of Guelph (School of Engineering, Department of Biomedical Engineering)* Made a significant contribution to the study design (specifically with the impact testing equipment), a minimal contribution to the study execution, a minimal contribution to data analysis and interpretation and provided guidance on manuscript preparation.

Timothy Burkhart - *The University of Western Ontario (School of Engineering, Department of Biomedical Engineering)* Made a minimal contribution to study design, a minimal contribution to study execution, a significant contribution to data analysis and interpretation (acceleration data) and provided significant guidance on manuscript preparation.

Warren Bignell - *The University of Guelph (Department of Biomedical Science)* Made a significant contribution to the study design (specifically with building of impact testing equipment and data acquisition system) and a significant contribution to study execution.

**Keywords for publication** – horse, biomechanics, impact, hoof, acceleration, metacarpophalangeal

**Declarations:**

Legal and ethical requirements - An ethical committee review was not required as this study used cadaver limbs that were severed post-mortem from animals that has either died of natural causes or had been euthanized for some other reason unrelated to this study. Limbs were obtained only after the owner or affiliation had given signed consent to donate the carcass.

Competing Interests – there are none

Source of Funding – NSERC

Acknowledgements - Thank you to William Sears for statistical analysis consulting.
2.1 Abbreviations

P1 – First phalanx

MC3 – Third metacarpal bone

P2 – Second phalanx

ANOVA – Analysis of variance

Pkacc - Peak deceleration following first contact

Ttp - Time from first contact to that peak

Pkfreq - Vibration frequency with maximal amplitude

En95 - 95% of the total energy in the signal

Freq95 - Frequency at 95% energy

s.d – Standard deviation

D-P - Dorsopalmar

CoV – Coefficient of variation

MCP – Metacarpophalangeal joint
2.2 Abstract

**Reasons for performing study:** We wished to add to the existing baseline data on impact loading of the distal limb as a precursor to assessing the potential role of impact in injury and joint disease.

**Objectives:** To examine the effect of three hoof-strike conditions (toe-first, flat, heel-first) and two specimen masses (with and without a ballast of approx. 2% body mass) on impact deceleration and vibration frequencies and energies at the hoof, first phalanx (P1) and third metacarpal (MC3).

**Study Design:** Eight cadaver limbs were subjected to randomized, repeated controlled trials in which the hoof was struck by a pendulum impact testing machine (impact velocity 3.55m/s) under the three strike and two mass conditions

**Methods:** Data from three-axis accelerometers on the hoof, P1 and MC3 quantified, for all trials, the peak impact acceleration, frequencies in the first 6.4ms following impact, the frequency with the most energy, 95% of the total energy, and the frequency at 95% cumulative energy. The effects of the strike and mass conditions on each variable were statistically tested using repeated-measures ANOVA (α=0.05).

**Results:** Signal energy reaching MC3 was 6-31% of that at the hoof. A heel-first strike produced the largest peak accelerations and highest frequencies among all strike conditions, and changing the mass had no effect regardless of strike condition.

**Conclusions:** Large accelerations that occur upon impact of the hoof with the ground are attenuated by the distal structures of the equine limb, but still carry considerable energy within
the signal that could be damaging to tissue and are dependent on hoof strike condition but not ballast.

**Potential relevance:** Despite significant signal dampening by the hoof, impact induces high magnitude shocks that reach up into MC3 and should be considered for future study in the context of injury.

### 2.3 Introduction

In the context of the causation of injury in the distal limb, loading at the moment of impact with the ground (Benoit et al. 1993, Gustås et al. 2006, Willeman et al. 1999, Lanovaz et al. 1998) has received considerably less attention than at midstance (Les et al. 1997, Nunamaker et al. 1990). Force magnitudes are considerably lower at impact (approx.2-10% of the peak at midstance), but it is known that repetitive impact can be involved in the etiology of osteoarthritis (Gustås et al. 2006, Radin et al. 1972, 1984, Serink et al. 1977). Previous studies, both *ex vivo* and *in vivo* have shown that the magnitude of deceleration upon impact is dependent upon impact velocity and is attenuated by approximately 90% between the hoof and metacarpus (Gustås et al. 2006, Willemen et al. 1999, Lanovaz et al. 1998). While these studies have shown significant attenuation from hoof to metacarpus, the impact velocity used was significantly less (0.5-1.8 m/s) than would be experienced by a horse during a regular training session (3.0-7.2 m/s).

Among the factors that are likely to affect impact are the material of the contact surface, the manner of contact and the effective mass at contact. A well-balanced hoof should land flat at the moment of impact (O’Grady et al. 2003), but toe first and heel first landings are other strike
possibilities with heel first contact being the most prevalent (Clayton et al. 1990, Van Heel et al. 2005), depending on conformation, speed and footing surface (Clayton et al. 1994, Back et al. 1995, Linford et al. 1994). Manipulation of toe angle of the hoof and pastern angle can affect the manner of loading (Clayton et al. 1990). Effective mass is defined as the fraction of the horse’s body mass that is involved at impact. Owing to the high accelerations, 590-3900 m/s² in Standardbreds (Chataeu et al. 2010) and 100-1050 m/s² in Thoroughbreds (Ratzlaff et al. 2005, Thomason et al. 2008, Thomason et al. 2007), it is important to keep the effective mass low in order to avoid dangerously high forces at impact.

This ex vivo study builds on previous studies of impact and vibration transmission in the distal limb. Its purpose is to consider the effects of landing type and effective mass on peak impact deceleration, the frequencies and energies in the subsequent vibration, and the attenuation from the hoof to the third metacarpal bone (MC3).

2.4 Materials and Methods

2.4.1 Specimens

Eight unshod Standardbred forelimbs (2 pairs, 2 isolated rights, 2 lefts) were transected 15 cm above the carpus, post mortem, from 6 horses that were subject to euthanasia for reasons other than pathologies of the locomotor system. The limbs were frozen (-20 °C) immediately after transection and allowed to thaw in a cooler for 12-16 h prior to mechanical testing.

2.4.2 Specimen preparation

Three triaxial accelerometers (Model 356A03, PCB Piezoelectronics, Depew, New York) were attached to the dorsum of MC3 and the first phalanx (P1) using custom bone screws (8mm
diameter, 4mm long), and to the dorsum of the hoof via a baseplate attached with epoxy (Equi-Thané Super-Fast foal shoe glue, Vettec Hoof Care) (Figure 2.8.1). Minimal soft tissue was removed to access the bone locations and no tendons were severed. Once mounted, the base of each accelerometer was flush with the bone’s surface. Each specimen was suspended by a pair of V-shaped cables from a custom designed jig that was attached to a pendulum impact device (Figure 2.8.1), and aligned to closely simulate primary impact according to kinematic data collected in vivo (Clayton et al. 1994, Butcher et al. 2002). Tensioners in the suspension cables allowed precise alignment of the sole of the hoof against the impactor head.

2.4.3 Testing apparatus

A single hammer Universal Pendulum Impact Tester (total mass 29.5 kg; Tinius Olsen – Horsham, PA, USA), faced with a ¼ inch industrial-strength geotextile and vulcanized rubber stall matting (StableComfort top cover, Promat Inc., Woodstock, ON) (Figure 2.8.1) was used for this experiment. A trigger allowed for a constant velocity of 3.55 m/s at contact. The device had an internal mechanism to stop the pendulum at contact, preventing multiple impacts. This apparatus, therefore, reproduced the primary impact phase of the stance without subsequent loading that occurs in secondary impact. This design did not accommodate for multiple contact surfaces or surface properties, but provided a consistent and repeatable surface for testing. It brought the impact accelerations within the range of those measured in trotting horses on a standard North American Standardbred racetrack (Thomason et al. 2008, Salo et al. 2010), and gave us the ability to compare strike and ballast conditions.

2.4.4 Accelerometers
Accelerometers (capacity, ±5000 m/s²; resonant frequency, 5500 Hz; frequency range, 0-4500 Hz) were aligned with the x-axis running proximo-distal along the longitudinal axis of the forelimb, the y-axis latero-medially and the z-axis dorso-palmarly (Figure 2.8.1). Accelerometer data were collected using an 8-channel A-D system (Logbook 300, Iotech Inc., Cleveland, OH) at sampling rate of 11 kHz. To record all nine inputs, the y channel inputs for P1 and MC3 were alternated between trials.

2.4.5 Hoof angle at contact and limb ballast

The strike angle of the solar surface to the impactor was manipulated among trials, by means of slight adjustments to the lengths of the suspending cables, and was measured from photographs taken from a standard lateral location.

An 11.4-kg ballast weight (approximately doubling the mass of each specimen) was secured with circular clamps to the proximal end of the limb (Figure 2.8.1) for three trials under a flat hoof strike to examine the effect of changing effective mass on primary impact. A bracket extending around the proximal end of the carpus ensured that the ballast was rigidly linked to the specimen in the x direction.

2.4.6 Testing Protocol

Fifteen randomized trials were performed for each specimen: six flat, three toe strike, three heel strike, and three flat with ballast. The impact velocity of 3.55 m/s was greater than in previous acceleration studies (Gustås et al. 2006, Willemen et al. 1999, Lanovaz et al. 1998) but is well within the normal in vivo range: 1.43 m/s for a slow trot to 7.2 m/s for a racing trot (Lanovaz et al. 1998, Gustås et al. 2001).
2.4.7 Data Analysis

The data were reduced using a custom written Matlab program, which isolated the first 6.4 milliseconds after first contact. A low-pass 9th-order Butterworth filter was applied, at a cut-off of 4000 Hz. The following variables were extracted for each accelerometer location and axis: peak deceleration following first contact (pkacc); time from first contact to that peak (ttp); vibration frequency with maximal amplitude, i.e., energy, (pkfreq); 95% of the total energy in the signal (en95), and the frequency at 95% energy (freq95), which indicates the upper range of significant vibration frequencies. A cubic spline was used for extrapolating the peak acceleration (pkacc_x) in a few heel strike trials (23 trials out of 120) in which the peak slightly overshot the range of the accelerometer. The extremely short sampling duration limited the resolution of the two frequency variables (pkfreq and freq95) by fast Fourier analysis. Frequencies were lumped within bins of width 156.25 Hz.

2.4.8 Statistical Analysis

Statistical tests were performed in SAS 9.3 (Statistical Analysis Services, Carey, NC). A repeated-measure ANOVA included horse as a random effect and strike angle and ballast as fixed covariates. After preliminary tests for normality, a logarithmic transformation was applied to each variable. Least mean-square differences were used for pairwise comparisons (α level = 0.05) between strike and ballast effects using median values (because of the transformation) and 95% confidence intervals were reported.

2.5 Results

2.5.1 Alignment of the three accelerometers
The average differences in x-axis orientation between the accelerometers mounted on the bones versus that on the hoof was 5.36° for P1 and 8.76° for MC3. These small angles were not accounted for in the data analysis, as they were consistent among loading conditions.

2.5.2 Angle of hoof strike with respect to the impactor

The mean angles between the sole of the hoof and the impactor head at the moment of impact under the three impact conditions were: flat strike 0°; heel-strike mean 4.44° (s.d +/- 1.17°, range 2.0-6.9°), toe-strike mean 3.54° (s.d +/- 0.84°, range 2.1-5.2°).

2.5.3 Impact magnitude and timing axially along the limb (x-axis)

Median peak acceleration (pkacc_x) at the hoof was in a range of 3058 – 5405 m/s², depending on manner of strike, and fell to 31-44% of hoof values at P1 and to 28-37% at MC3 (Figure 2.8.2). Most of the comparisons among strike condition were significant (p<0.05) at each accelerometer location (within group), as were comparisons across locations (between groups).

Median times to peak acceleration (ttp_x) increased significantly (p<0.05) from hoof to both P1 and MC3, for each strike condition, with non-significant increases from P1 to MC3 (Figure 2.8.3). Heel strikes resulted in significantly longer times to peak than flat or toe strikes (p<0.05), at all three locations, while flat and toe strikes gave significantly different times only at the hoof.

2.5.4 Frequency and energy of vibration along the x-axis
The highest peak frequencies (pkfreq) were found at the hoof and were similar under all hoof strike conditions ranging from 157-1250 Hz (Figure 2.8.4). Mean peak frequency was reduced to 16% at P1 and 15% at MC3 of that measured at the hoof.

The highest frequencies (freq95_x) were found at the hoof for location and during a heel strike for condition (Figure 2.8.5). Energy in the signal (en95_x) was greatest during a heel strike in the hoof, a normal strike at P1 and a toe strike at MC3 (Figure 2.8.6). Energy levels at the hoof were reduced to 6-31% at MC3. Pairwise testing indicated there was a significant difference between all three locations under the same hoof strike with some significance between hoof strike at the same location (Figure 2.8.7).

2.5.5 Results in transverse (y) and D-P (z) directions

The means for peak acceleration (pkacc) data in the y-direction and the z-direction were lower when compared to the x-direction and the data had greater variability (higher coefficients of variation, CoV) in the y and z-direction compared to the x-direction (Table 2.1). The means for frequency (freq95) in the y-direction were the greatest compared to the x and z-directions, while the means for the energy in the signal (en95) were lower in the y and z-directions compared to the x-direction (Table 2.1).

2.5.6 Changing the effective mass

Addition of the ballast did not have a significant effect (p>0.05) on any of the variables tested (Table 2.2).

2.6 Discussion
The purpose of this study was to examine the effect of hoof strike and the effective mass on time- and frequency-domain measures of impact. We expected that hoof strike and ballast would have a significant effect on the accelerations measured at all three locations, which was true for most comparisons involving hoof strike, but not for ballast.

2.6.1 Study Limitations

Failure to preload the flexor and extensor tendons in the distal forelimb may have created a source of error within the experimental set-up. It has been shown that although tension in the digital flexors and extensors does exist prior to hoof colliding with the ground, there is almost no rotation of the MCP joint at that time suggesting that the forces applied are equivalent but opposing (Harrison 2012). Since primary impact occurs more quickly than the neuromuscular response time (50ms), the tendons were considered to be acting passively at this phase of the stance (Lanovaz et al. 1998). The overall design of the testing jig had limitations in the ability to replicate an exact foot fall of a live animal. Although it did not accommodate for multiple surface properties, it did provide a consistent and repeatable testing surface that upon impact brought the accelerations within a similar range to those measured in trotting horses on a standard North American Standardbred racetrack (Thomason et al. 2008, Salo et al. 2010). Our aim for this study was to measure the accelerations occurring at primary impact under multiple hoof strike conditions, the jig design allowed us to isolate primary impact without confounding our results due to subsequent loading. Although loading due to body weight occurs in the live animal, it was not considered as an important biomechanical component to the objective of this study.

2.6.2 Peak Acceleration
Previous studies in vitro (Lanovaz et al. 2006) and in vivo (Gustás et al. 2001) recorded peak accelerations at the hoof that were comparable to our findings in this study. The accelerations measured at the hoof (3735 – 4911 m/s²) are similar to those found in the live horse moving at higher speeds (Gustás et al. 2001). Peak accelerations previously reported (Willemen et al. 1999) for P1 and MC3 were considerably lower than the accelerations measured in this study (1174 - 2287 m/s²) due to the increase in impact velocity used in the current study. Across all sites, a heel strike produced the highest peak accelerations, possibly due to the rotational acceleration of the toe down onto the plate after heel contact.

2.6.3 Time to Peak

The first permanent departure from the noise around signal zero in any channel indicated the time of impact. For a toe and flat strike we expected the same delay between actual contact and ttp as indicated by our results (Figure 2.8.3). The ttp for a heel strike was longer due to the additional distance the signal travelled before reaching the accelerometer and travelling through the softer heel tissue, damping the signal before it reached its peak acceleration. Heel strike also has two contact times: one when the heel hits with a relatively low peak acceleration and another later when the toe rotates around the heel and makes contact itself, with higher peak acceleration. This second peak was extracted for the present data set, hence the longer ttp, but this manner of contact explains why there is a similar difference between hoof, P1 and MC3 under all strike conditions.

2.6.4 Peak Frequency and Frequency at 95%

Hoof strike had no significant effect on vibration frequency however the results showed high frequencies being experienced at all locations. Although high-frequency vibrations are
thought to contribute to bone injury (Gustás et al. 2006, Radin et al. 1972, Serink et al. 1977, Hjerten et al. 1994), there is little evidence that has determined a frequency threshold associated with bone damage and failure. However, the anabolic effect that high-frequency, low-magnitude vibration has on bone suggests that frequency is a primary stimulus for bone adaptation, opposed to force magnitude (associated with strain) (Judex et al. 2007, Garman et al. 2007). While there is an abundance of literature that exists on biomechanical quasi-static loading of bone (loading associated with low frequencies – under 10Hz – and high magnitudes) and bone strain, there are only a few studies, either in vitro (Willemen et al. 1999, Lanovaz et al. 1998) or in vivo (Benoit et al. 1993, Gustás et al. 2006), that have measured bone frequency (474 – 1787 Hz) associated with impact in the horse depending on location. Overall, our results were similar to the previous findings (208 – 1235 Hz) indicating that the hoof is responsible for absorbing and attenuating a large percentage of the high frequencies at the moment of impact however, there is still significant energy reaching MC3 despite this damping.

2.6.5 Energy at 95%

Overall, the greatest reduction in total energy across the accelerometer locations occurred during a heel strike agreeing with the suggestion that the heel of the hoof acts as an effective shock absorber due to the soft tissues located in this area (Thomason et al. 2008). Although a heel strike had the greatest energy reduction, it also produced the highest frequency and greatest total energy at the hoof compared to all other locations and strike conditions. A toe strike produced the lowest decrease in energy of the signal (due to the lack of heel shock absorption); however it had lower peak accelerations and lower frequencies (freq95) in the signal. The data from Lanovaz et al. 1998 suggests that much of the amplitude attenuation (energy absorption) occurs across the proximal interphalangeal joint. We were unable to comment on this finding as
we did not place accelerometers on both P1 and P2. Our findings indicate that while there was no significant frequency attenuation, there was significant energy reduction across the MCP joint from P1 to MC3 (Figure 2.8.6) suggesting that the MCP joint is absorbing considerable energy, which could be a factor in contributing to bone injury.

2.6.6 *Peak Acceleration Transmission*

Peak acceleration at MC3 has been reported as being as low as 10% of the original magnitude at the hoof (Willemen et al. 1999). Our results show considerably higher percentages of transmission: 31-44% and 28-36% at P1 and MC3 respectively from that measured at the hoof. It has been suggested that the soft tissues in the distal limb such as the deep digital flexor tendon, are primarily responsible for damping the high-frequency vibrations after impact (Wilson et al. 2001). However, primary impact does not engage these soft tissues and therefore does not act on the springing system (Thomason et al. 2008). While frequencies at P1 and MC3 have been damped considerably from the frequencies measured at the hoof, it is possible that they could still be considered damaging to bone, however further investigation in this area is required.

2.6.7 *Effective Mass*

It is difficult to determine the effective mass that is involved during primary impact in the horse. Even with an addition of 2% of the total animal’s body weight, our study was unable to detect any changes among the variables measured.

These findings suggest that shock compression waves travelling from the hoof did not reach the proximal part of the limb during primary impact, and did not engage the ballast during the 6 ms analysis time frame.

2.6.8 *Farriery and Bone Injury Implications*
It is common practice to trim the hooves of racehorses to encourage them to land heel first at the moment of impact (Clayton et al. 1990). Based on our results, a heel strike is associated with the highest peak accelerations and peak frequencies. Although it is unknown if these signals are potentially dangerous, further investigation may find that trimming to encourage this type of impact could have negative implications and contribute the increase in SCB stiffening as seen in the distal end of MC3 in horses with MCP OA.

Although there is a lack of published evidence that links high energy physiological impact loading to bone damage, it is possible that the energy absorption occurring through the MCP joint could contribute to bone injury within the joint. Approximately one-third of young Standardbred racehorses have osteoarthritis within the MCP joint with prevalence increasing with age (Neundorf et al. 2010). It has been suggested that high-frequency vibrations orchestrate a bone response that produces a stiffer, thicker trabecular structure (Ocvivici et al. 2007). Thickening of the subchondral bone plate and endochondral ossification leads to tidemark advancement towards the joint surface effectively reducing the thickness of the articular cartilage and increasing the stresses at the base of the overlying cartilage (Burr and Radin 2003). Analysis of subchondral bone within the MCP joint of racehorses has shown increased bone density in certain areas of MC3 causing an increase in bone stiffness in response to mechanical loading (Rubio-Martinez et al. 2008, Drum et al. 2008, Muir et al. 2008).

2.7 Conclusions

Large accelerations and high-frequency vibrations upon impact of the hoof with the ground are modulated and attenuated by the distal structures of the equine limb. There is still considerable energy in the peak magnitudes of the impact signal reaching MC3, which could be
damaging to tissue and is dependent on hoof strike condition but not small changes in effective mass.
2.8 Figures and Tables

Figure 2.8.1 - Experimental apparatus used to simulate primary impact. Dark line on hammer head face represents the rubberized contact surface.
Figure 2.8.2 – Peak acceleration in the x-axis (pkacc_x) following contact, at all locations (hoof, P1, MC3) and for all hoof-strike conditions (heel, flat, toe). Bars sharing the same letter below each chart did not show statistical significance (p<0.05). Numbers above the P1 and MC3 bars are percentages of the equivalent hoof bar. Insert shows representative acceleration traces from one impact: red, hoof; blue, P1, and green, MC3. The signal in the inset deviated from zero approx. 3 ms after the signal from one y axis accelerometer, hence the apparent lag after time zero.
Figure 2.8.3 – Time to peak in the x-axis (ttp_x), by location and strike. The solid bars represent the mean and the standard error of the mean is indicated by the line above and below the mean. Bars sharing the same letter below each chart did not show statistical significance (p<0.05).
Figure 2.8.4 – Frequency with maximal energy (pkfreq) by location and hoof strike. Reported are the pkfreq_x means and the frequency range. A range was given based on the width of the collection data bins as detected from the fast Fourier transform. The only significant difference was found between a heel strike and a toe strike at the MC3 location.
**Figure 2.8.5** – Frequency below which 95% of the cumulative energy is found (freq95), by location and hoof strike. The solid bars represent the mean and the standard error of the mean is indicated by the line above and below the mean. Bars sharing the same letter below each chart did not show statistical significance (p<0.05).
**Figure 2.8.6** — Cumulative energy below the 95% frequency (en95), by location and strike. The solid bars represent the mean and the standard error of the mean is indicated by the line above and below the mean. Bars sharing the same letter below each chart did not show statistical significance (p<0.05).
<table>
<thead>
<tr>
<th></th>
<th>X Mean (CoV)</th>
<th>Y Mean (CoV)</th>
<th>Z Mean (CoV)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Pkacc (m/s²)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Heel</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>4911 (0.20)</td>
<td>190 (2.85)</td>
<td>290.7 (1.80)</td>
</tr>
<tr>
<td>P1</td>
<td>2287 (0.36)</td>
<td>224 (3.84)</td>
<td>1351 (1.09)</td>
</tr>
<tr>
<td>MC3</td>
<td>1793 (0.46)</td>
<td>383 (1.71)</td>
<td>1178 (0.67)</td>
</tr>
<tr>
<td><strong>Flat</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>3735 (0.30)</td>
<td>5.18 (74.80)</td>
<td>332.3 (1.97)</td>
</tr>
<tr>
<td>P1</td>
<td>1650 (0.34)</td>
<td>316 (3.08)</td>
<td>1325 (0.43)</td>
</tr>
<tr>
<td>MC3</td>
<td>1372 (0.28)</td>
<td>132 (5.14)</td>
<td>956.4 (0.57)</td>
</tr>
<tr>
<td><strong>Toe</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>3786 (0.18)</td>
<td>57.4 (10.10)</td>
<td>354.2 (1.53)</td>
</tr>
<tr>
<td>P1</td>
<td>1174 (0.25)</td>
<td>443 (0.92)</td>
<td>1019 (0.46)</td>
</tr>
<tr>
<td>MC3</td>
<td>1068 (0.29)</td>
<td>328 (0.93)</td>
<td>615.1 (0.57)</td>
</tr>
<tr>
<td><strong>Freq95 (Hz)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Heel</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>1235 (0.53)</td>
<td>1414 (0.61)</td>
<td>1035 (0.73)</td>
</tr>
<tr>
<td>P1</td>
<td>703 (0.96)</td>
<td>1488 (0.47)</td>
<td>710.9 (0.34)</td>
</tr>
<tr>
<td>MC3</td>
<td>803 (0.66)</td>
<td>930.9 (0.43)</td>
<td>669.6 (0.37)</td>
</tr>
<tr>
<td><strong>Flat</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>810 (0.19)</td>
<td>937.5 (0.33)</td>
<td>515.2 (0.38)</td>
</tr>
<tr>
<td>P1</td>
<td>797 (1.03)</td>
<td>1026 (0.23)</td>
<td>669.6 (0.60)</td>
</tr>
<tr>
<td>MC3</td>
<td>338 (0.52)</td>
<td>1635 (0.72)</td>
<td>726.9 (0.38)</td>
</tr>
<tr>
<td><strong>Toe</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>726 (0.20)</td>
<td>2260 (0.50)</td>
<td>460.9 (0.32)</td>
</tr>
<tr>
<td>P1</td>
<td>208 (0.61)</td>
<td>1367 (0.51)</td>
<td>558.0 (0.15)</td>
</tr>
<tr>
<td>MC3</td>
<td>323 (1.31)</td>
<td>1283 (0.62)</td>
<td>750.0 (0.77)</td>
</tr>
<tr>
<td><strong>En95 (m/s²<em>Hz²</em>10⁶)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Heel</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>106 (0.45)</td>
<td>0.99 (2.30)</td>
<td>65.8 (0.35)</td>
</tr>
<tr>
<td>P1</td>
<td>36.9 (0.61)</td>
<td>6.14 (1.02)</td>
<td>29.5 (1.21)</td>
</tr>
<tr>
<td>MC3</td>
<td>13.6 (0.66)</td>
<td>6.02 (1.28)</td>
<td>16.0 (0.86)</td>
</tr>
<tr>
<td><strong>Flat</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>71.7 (0.54)</td>
<td>1.10 (2.44)</td>
<td>51.2 (0.38)</td>
</tr>
<tr>
<td>P1</td>
<td>44.5 (0.25)</td>
<td>7.10 (0.95)</td>
<td>21.0 (0.59)</td>
</tr>
<tr>
<td>MC3</td>
<td>14.9 (0.30)</td>
<td>3.70 (1.14)</td>
<td>12.4 (0.83)</td>
</tr>
<tr>
<td><strong>Toe</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>53.5 (0.40)</td>
<td>1.85 (1.89)</td>
<td>25.7 (0.45)</td>
</tr>
<tr>
<td>P1</td>
<td>32.1 (0.26)</td>
<td>6.32 (0.41)</td>
<td>13.1 (0.55)</td>
</tr>
<tr>
<td>MC3</td>
<td>15.9 (0.31)</td>
<td>2.42 (1.15)</td>
<td>11.4 (1.15)</td>
</tr>
</tbody>
</table>

**Table 2.1** - Values comparing means and coefficients of variation of trials without ballast under all strike conditions at all locations, and in all directions. The large variability among these values is likely due to the off axis angle the accelerometers had in relation to each other.
<table>
<thead>
<tr>
<th></th>
<th>Ballast</th>
<th>No Ballast</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>pkace_x (m/s²)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>3800 (4436-3255)</td>
<td>3569 (4166-3057)</td>
<td>0.5390</td>
</tr>
<tr>
<td>P1</td>
<td>1278 (1651-1212)</td>
<td>1572.22 (1835-1346)</td>
<td>0.3037</td>
</tr>
<tr>
<td>MC3</td>
<td>1414 (1492-1095)</td>
<td>1319.37 (1540-1130)</td>
<td>0.7599</td>
</tr>
<tr>
<td><strong>ttp_x (ms)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>0.82 (0.905-0.735)</td>
<td>0.86 (0.920 -0.820)</td>
<td>0.2388</td>
</tr>
<tr>
<td>P1</td>
<td>1.04 (1.155 -0.938)</td>
<td>1.07 (1.138 -1.014)</td>
<td>0.5561</td>
</tr>
<tr>
<td>MC3</td>
<td>1.12 (1.237 -1.006)</td>
<td>1.13 (1.120 -1.070)</td>
<td>0.7761</td>
</tr>
<tr>
<td><strong>freq95_x (Hz)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>787 (973.9 - 637.4)</td>
<td>790 (884.1 - 706.9)</td>
<td>0.9748</td>
</tr>
<tr>
<td>P1</td>
<td>201 (424.8 - 95.1)</td>
<td>319 (462.6 - 220.7)</td>
<td>0.2492</td>
</tr>
<tr>
<td>MC3</td>
<td>331 (572.5 - 192.4)</td>
<td>323 (419.7 - 249.07)</td>
<td>0.9290</td>
</tr>
<tr>
<td><strong>en95_x (m/s²Hz¹*10⁶)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hoof</td>
<td>69.5 (97.0 – 49.7)</td>
<td>69.7(86.5 – 56.1)</td>
<td>0.9860</td>
</tr>
<tr>
<td>P1</td>
<td>31.0 (79.8 – 15.7)</td>
<td>32.7 (86.8 - 13.6)</td>
<td>0.7400</td>
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<tr>
<td>MC3</td>
<td>11.2 (43.5 – 2.21)</td>
<td>10.8 (40.8 – 2.62)</td>
<td>0.8394</td>
</tr>
</tbody>
</table>

**Table 2.2** – Values comparing median values and confidence intervals of trials with ballast compared to those without ballast under a flat strike condition. Significance set at (P<0.05).
CHAPTER 3: EFFECT OF HOOF ANGLE ON JOINT CONTACT AREA IN THE EQUINE METACARPOPHALANGEAL JOINT FOLLOWING SIMULATED IMPACT LOADING EX VIVO

Accepted for publication to the Equine Veterinary Journal

Authors and Contributions:

Cristin A. McCarty – The University of Guelph (Department of Biomedical Science) The main contributor to study design, did the majority of the study execution, the majority of the data analysis and interpretation, and the majority of the manuscript preparation.

Jeffrey J. Thomason - The University of Guelph (Department of Biomedical Science) Made a significant contribution to the study design, a small contribution to the study execution, a significant contribution to the data analysis and interpretation (programming in Matlab) and provided guidance on manuscript preparation.

Karen Gordon - The University of Guelph (School of Engineering, Department of Biomedical Engineering) Made a significant contribution to the study design (specifically with the impact testing equipment), a minimal contribution to data analysis and interpretation and provided guidance on manuscript preparation.

Mark Hurtig - The University of Guelph (Department of Clinical Studies) Made a significant contribution to study design and provided guidance on manuscript preparation.

Warren Bignell - The University of Guelph (Department of Biomedical Science) Made a significant contribution to the study design (specifically with building of impact testing equipment) and a significant contribution to study execution.

Corresponding author’s email address: Jeffrey J Thomason (jthomaso@ovc.uoguelph.ca)

Keywords for publication – horse, biomechanics, impact, metacarpophalangeal, joint loading, osteoarthritis

Word count – 4199 words

Declarations:

Legal and ethical requirements - An ethical committee review was not required as this study used cadaver limbs that were severed post-mortem from animals that has either died of natural causes or had been euthanized for some other reason unrelated to this study. Limbs were obtained only after the owner or affiliation had given signed consent to donate the carcass.

Competing Interests – there are none

Source of Funding – Natural Sciences and Engineering Research Council of Canada (NSERC)

Acknowledgements - Thank you to William Sears for statistical analysis consulting.
3.1 Abbreviations

MCP – Metacarpophalangeal joint

MC3 – Third metacarpal bone

P1 – First phalanx

PS – Proximal sesamoids

OA – Osteoarthritis

S – Scaling factor

A – Total high-pressure contact area total areas (cm²)

n – Number of pixels above limit

Cₓ - Centroid location between the centroids and the sagittal ridge (cm)

Cᵧ - Centroid location between the centroids and the transverse ridge (cm)

Iₓᵧₙ - Dispersion of pixels within each area relative to the centroid (cm⁴)

d - Hypotenuse distance between each pixel and the centroid (cm)

ANOVA – Analysis of variance

CI – Confidence interval
3.2 Abstract

**Reasons for the Study**: To add to the existing baseline data on impact loading of the metacarpophalangeal (MCP) joint as a precursor to assessing the potential role of impact in injury and joint disease.

**Objectives**: To examine the effect of impact loading on relative contact areas of the first phalanx (P1) and proximal sesamoid bones (PS) with the third metacarpal (MC3) under three hoof-strike conditions (toe-first, flat, heel-first).

**Study Design**: Eight cadaver limbs were subjected to randomized, repeated controlled trials in which the hoof was struck by a pendulum impact machine (impact velocity 3.55m/s) under the three strike conditions.

**Methods**: Data from pressure sensitive film placed over medial and lateral MC3 condyles and latero-medially across the dorsal aspect of MC3 were quantified: total areas of P1 and PS contact (A, cm²) at maximum recorded pressure; centroid locations of contact areas relative to the sagittal ridge (Cx, cm) and transverse ridge (Cy, cm), and dispersion of pixels (I_{xy,n}, cm⁴) for each MC3 condyle (medial/lateral). The effect of the strike conditions on each variable were statistically tested using repeated-measures ANOVA (α=0.05).

**Results**: Contact area between P1 and MC3 condyles fell in well-defined areas bounded by the sagittal and transverse ridge, whereas contact areas from PS were smaller and widely dispersed across MC3 palmar border. The ratio of contact area of P1 to PS was 2.83 (P < .0001). Hoof strike had no significant effect on contact area (p>0.54)
Conclusions: Contact at impact (primarily from P1 and distally situated on MC3), contrasts with contact areas at midstance from both P1 and PS, symmetrically placed.

Relevance: Under impact, the greatest contact area was on the dorsal aspect of the medial condyle and coincides with the area subject to the greatest increase in subchondral bone stiffening in joint disease.

3.3 Introduction

Osteoarthritis (OA) in the metacarpophalangeal joint (MCP) of racehorses is accompanied by changes in bone architecture and joint geometry that increase as the disease progresses (Easton et al. 2007, Drum et al. 2008, Muir et al. 1993). Exercise-induced bone remodeling in the condylar region of the third metacarpal (MC3) can cause subchondral bone sclerosis and damage to the articular cartilage, leading to degenerative lesions in advanced OA. These lesions weaken the joint integrity and can lead to catastrophic condylar failure that ultimately results in early retirement or euthanasia of young horses (Parkin et al. 2006, Whitton et al. 2010).

The architectural changes observed with MCP OA are a response to mechanical loading sustained during high-speed racing and training. The unique joint configuration, high range of motion and small surface area provide multiple loading sites as the joint moves from flexion into extension. Contact stress (the ratio of the force magnitude over the contact area) during midstance (i.e., at full joint extension) is associated with site-specific changes within the distal end of MC3 (Young et al. 2007, Riggs et al. 1999). It has been reported that the palmar aspect of MC3 is most susceptible to these changes (Riggs et al. 1999, Rubio-Martinez et al. 2008),
suggestive of high midstance pressure from the proximal sesamoids (PS). New evidence from micro-imaging, however, suggests that both palmar and dorsal aspects of the medial condyle show changes in bone micro-architecture among young racehorses (Young et al. 2007). Previous modelling of the mechanics of the joint shows that the distal condyle of MC3 is compressed, between P1 and PS at midstance, with high dorsal and palmar loading (Easton et al. 2007).

Experimental data have shown that joint pressure within the MCP joint peaks during midstance, and increases with the forward velocity of the horse (Easton et al. 2007, Brama et al. 2001, Hartog et al. 2009). Contact area of the proximal phalanx (P1) on MC3 also increases with velocity, and shifts dorsally (Easton et al. 2007, Brama & al. 2001, Hartog et al. 2009) as a result of increased joint extension (Clayton et al. 2007). Increased stiffness in subchondral bone at this contact area is thought to result from the high forces that occur during midstance (Easton et al. 2007, Brama et al. 2001, Colahan et al. 1988). It is not known if loading or the contact area at other phases during the stance may also affect bone remodelling. We have begun to examine whether the loading conditions at primary impact (the few milliseconds after hoof contact with the ground) may also have potentially deleterious effects (McCarty et al. 2014).

High vertical accelerations measured during impact loading (high magnitude, high frequency) have also been shown to elicit bone changes and contribute to damage within a joint (Radin et al. 1972). Repetitive impact loading under high speed locomotion that occurs during racing and training imposes large stresses on the metacarpophalangeal (MCP) joint which over time could cause stiffening of the subchondral bone initiating a cascade of events that eventually lead to the thinning of the articular cartilage and increased risk of shear stress to the overlying cartilage (Burr and Radin 2003). Factors likely to affect impact loading include hoof-surface interaction, impact velocity and hoof orientation as it makes contact with the ground. A well-
balanced hoof should land flat at the moment of impact (O’Grady et al. 2003), but toe first and heel first landings are other strike possibilities with heel first contact being the most prevalent (Clayton et al. 1990, Van Heel et al. 2005), depending on conformation, speed and surface (Clayton et al. 1990, 1994, Van Heel et al. 2005). The purpose of this study was to determine the contact areas and their locations within the equine MCP joint under impact loading at multiple hoof strike conditions as a comparison to previous work that has characterized contact area and location during midstance loading (Easton et al. 2007, Brama et al. 2001, Hartog et al. 2009).

3.4 Materials and Methods

3.4.1 Specimens

Eight unshod Standardbred forelimbs (2 pairs, 2 isolated rights and 2 lefts) were transected 15 cm above the carpus, post mortem, from 6 horses (age 3-7 years) that were subject to euthanasia for reasons other than pathologies of the locomotor system. The limbs were frozen (-20 °C) immediately after transection and allowed to thaw in a cooler for 12-16 h prior to mechanical testing.

3.4.2 Specimen preparation

A transverse incision was made into the dorsal aspect of the MCP joint, approximately 1cm distal to the attachment of the capsule to MC3, and across most of the width of that bone. Care was taken to avoid severing the collateral ligaments during this process. Pressure sensitive film was later slid through the incision, between P1 and MC3 (details below). Each specimen was suspended from a custom made jig that was attached to a pendulum impact device (Figure 1). Two suspension cables were used, one near the MCP joint (small incisions were made to allow the cable to pass between the deep digital flexor tendon and palmar aspect of MC3 to
prevent interference across the joint) and the other proximal to the carpus (secured with a metal hose clamp with care taken to not apply tension to tendons). The limb was mounted in a manner that closely approximates the alignment at impact according to kinematic data collected in vivo (Back et al. 1995, Linford et al. 1994). This was achieved by mounting the limb with the dorsal aspect facing down in order to keep the limb straight (MCP joint angle between 170°- 175°), and by using adjustable tensioners within the supporting cables to vary the contact angle between hoof and the impact hammer.

3.4.3 Placement of intra-articular pressure sensitive film

Pressure sensitive film (Fujifilm Prescale®, USA, range 0.5 – 2.5MPa) was used to measure contact area and contact pressure at impact. The film shows different intensities of pigmentation to pressures within range, which can be assessed with a densitometric method (see Data Analysis). The films were cut according to the size and shape of the joint and were sealed within a transparent, polyethylene covering to prevent moisture from saturating the film. Films were placed over both the lateral and medial condyles on the distal end of the MC3 bone, spanning dorsal and palmar aspects of the bone to either side of the sagittal ridge (Figure 3.8.2). To assess contact on the ridge itself, films spanning the lateromedial width of the MCP joint were placed on the dorsal aspect of the MC3 – P1 contact area, in separate impact trials (Figure 3.8.2). Registration marks on the bone and film allowed their relative positions to be accurately determined Post hoc.

3.4.4 Testing apparatus and protocol

The impact device was a single hammer, Universal Pendulum Impact Tester (total mass 29.5 kg; Tinius Olsen – Horsham, PA, USA), faced with a ¼ inch industrial-strength geotextile
and vulcanized rubber stall matting (StableComfort top cover, Promat Inc., Woodstock, ON) (Fig 3.8.1). A trigger allowed for a constant velocity of 3.55 m/s at contact, which is within the normal in vivo range: 1.43 m/s for a slow trot to 7.2 m/s for a racing trot (Wilson et al. 2001, Lanovaz et al. 1998, Gustås et al. 2001). The device had an internal mechanism to stop the pendulum at contact, preventing multiple impacts. This apparatus, therefore, reproduced the primary impact phase of the stance without subsequent loading that occurs during secondary impact or midstance. Peak accelerations at impact were recorded from the hoof (3058 – 5405 m/s²), for a separate experiment conducted in parallel with this one (McCarty et al. 2014), and were within the range of those measured in trotting horses on a standard North American Standardbred racetrack (Thomason et al. 2008, Salo et al. 2010).

The strike angle of the solar surface of the hoof to the impactor was manipulated among trials, using the cable tensioners, to produce heel first, flat, and toe first strikes, in random sequence. The strike angle was measured for each trial from a photograph taken from a standard lateral location.

Films were introduced simultaneously on lateral and medial condyles of each specimen, and their location registered against a mark on the specimen. Four repeated strikes were conducted on the first 3 specimens to establish consistency. After that, each specimen was struck once for each of the 3 strike angles and the 2 film positions (both condyles together, or one lateromedially-oriented film). Fresh films were used for all strikes.

3.4.5 Data Analysis

Each film was scanned on a high-resolution flatbed scanner, adjacent to a visible scale, and the image imported into a custom written MATLAB program for scaling, and thresholding
into specified pressure levels (Figure 3.8.3). A scaling factor $S$ (cm) was calculated as the length of the scale divided by the number of pixels spanning it in the image, which gave the area ($\Delta A$) represented by each pixel as $S^2$. A selection of films were sent away for analysis on a calibrated densitometer (Sensor Products, Madison, NJ, USA), and these were used to validate the scans that we performed on all of the films.

Despite preliminary tests to establish an appropriate pressure range for the film, it was only found after all of the tests, that the upper end of the range was too low to capture the highest within-joint pressures. However, clear areas of contact were defined by the pixels showing pressure above the upper limit, outside of which the pressures dropped rapidly. We therefore extracted useful information on the areas and locations of the ‘over-limit’ areas (>100% of film range), rather than the pressure magnitudes within them.

Polygons were drawn on each image (in MATLAB) up to the borders of the transverse and sagittal ridges (Figure 3.8.4). These polygons completely surrounded areas that made contact, respectively, with P1 and PS. Within each area, pixels were identified within 6 levels of colour intensity on the film: Over limit, >100%; Level 4, 75 to <100%; Level 3, 50 to <75%; Level 2, 25 to <50%; Level 1, >0 to <25% and Background, 0%). All levels below 100% were used for qualitative description only (Figure 3).

The over-limit (>100%) level was used to identify total areas ($A$, cm$^2$) of high-pressure contact areas, as the number ($n$) of pixels above limit times $\Delta A$. Centroid location within the areas was calculated, using standard engineering formulae (Beer et al. 2000), as were the distances between the centroids and the sagittal ridge ($C_x$, cm) and the transverse ridge ($C_y$, cm) on each condyle. Lastly, a measure of dispersion of pixels within each area relative to the
centroid \( (I_{xy,n}, \text{cm}^4) \) was calculated as \( \sum \Delta A.d^2 / n \), where \( d \) is the hypotenuse distance between each pixel and the centroid. This method corresponds to measures of dispersion in many fields, including statistics (e.g., calculation of variance) and static mechanics (e.g., second moment of area) (Beer 2000). Dividing this quantity by \( n \) accounted for differences in total number of pixels in each contact area.

3.4.6 Statistical Analysis

Statistical tests were performed in SAS 9.3 (Statistical Analysis Services, Carey, NC). All statistical tests were performed on the over-limit data only. ANOVA tests were performed using horse as a random effect, with fixed covariates of: location (medial condyle or lateral condyle); bone (P1 or PS) and strike (heel, flat, or toe). After preliminary tests for normality, a logarithmic transformation was applied when necessary. Least mean-square differences were used for pairwise comparisons (\( \alpha \) level = 0.05) between strike effect using median values (because of the transformation), and 95% confidence intervals were reported where a logarithmic transformation was applied.

3.5 Results

3.5.1 Angle of Hoof Strike with Respect to the Impactor

The mean angles between the sole of the hoof and the impactor head at the moment of impact, for the three impact conditions were: heel strike mean 4.44° (s.d +/- 1.17°, range 2.0–6.9°); flat strikes 0°, and toe strike mean 3.54° (s.d +/- 0.84°, range 2.1–5.2°).

3.5.2 Contact Areas of the Proximal Phalanx (P1) and Proximal Sesamoid (PS) Bones
The area of contact with P1 was consistently, a single cohesive area, which lay adjacent to the sagittal and transverse ridges (Table 3.1, Figure 3.8.4). The contact area with PS was smaller in total area, and more widely dispersed across the palmar border of the MC3 articular surface (Figure 3.8.4). The lateromedial films consistently showed contact areas that extended across the sagittal ridge indicating a continuation of the area of contact from the lateral to medial condyle across the sagittal ridge.

The median total contact areas for P1 and PS were 1.91 cm$^2$ (95% CI 1.23 – 2.97cm$^2$) and 0.68 cm$^2$ (95% CI 0.43 – 1.06cm$^2$), respectively, when ignoring variation owing to hoof strike (Table 3.2). Area of contact with P1 was significantly higher than with PS (P <.0001) (Table 3.1) and the mean ratio of P1: PS contact area was 2.83.

3.5.3 Lateral and Medial Condyles Contact Area

The medial condyle had a greater total surface area compared to the lateral condyle, however both were found to be consistent, with a tightly clustered pixel dispersion.

The medians (and confidence intervals, CI) of total surface contact area for loc1 (lateral condyle) and loc2 (medial condyle) were 1.09 cm$^2$ (0.70 – 1.70 cm$^2$) and 1.19 cm$^2$ (0.76 – 1.85cm$^2$), respectively, over all trials (Table 3.2). The ratio of contact area loc1 to loc2 was 0.92 and was not statistically significant (p-value=0.53).

3.5.4 Contact Area by Strike

It appeared as though a toe strike was found to produce the greatest contact area, followed by a flat and heel strike (Figure 3.8.6), although none of the pairwise comparisons among them were statistically significant (p-values 0.58, 0.54, 0.69). The dispersion and
distribution of contact area among strikes were similar, with a minor dorsal shift in contact location between a heel and toe strike. The median total surface area for a heel, flat and toe strike was 0.97 (CI 0.62 – 1.52) cm$^2$, 1.22 (CI 0.77 – 1.94) cm$^2$ and 1.24 (CI 0.75 – 2.05) cm$^2$ (Table 3.2).

3.6 Discussion

The aims of the study were to measure contact areas of P1 and PS with the medial and lateral condyles of MC3 during simulated primary impact, under 3 hoof-strike conditions. Contact areas on MC3 differed significantly between contacting bones (P1 and PS), slightly by condyle location (lateral vs. medial), and slightly according to hoof strike. This is the first attempt to measure contact areas under impact loading for this joint.

3.6.1 Contact Location

Results from this study indicate that loading on MC3 during primary impact occurs primarily between P1 and the dorsal surface of MC3 as a result of the limb alignment that occurs during this phase of the stance (Figure 3.8.5). In general, the distribution of the loading from P1 was more condensed and clustered in the medial aspect of MC3 with loading occurring across the sagittal ridge compared to the loading from the proximal sesamoids which was widely distributed with random patterning. Greater loading from P1 compared to PS is plausible given that the angle of the MCP joint during primary impact has been measured to be between 165°-175° (Clayton et al. 1994, Thomason et al. 2008, Butcher and Ross 2002, Chateau et al. 2010). Unlike midstance loading in which high force loading with minimal acceleration occurs over two areas of contact (P1 and PS), impact loading generates lower forces, very high accelerations and occurs over a decreased total contact area (P1). Previous work has determined that as racing
speed increases, stress to the palmar aspect of the condyle was calculated to be more than twice of that applied to the dorsal surface (Riggs et al. 1999) indicating that there should be greater bone sclerosis on the palmar aspect of the condyle compared to the dorsal aspect. An in depth analysis of bone geometry and density changes using microCT within the distal MC3 of horses with OA showed that it was the dorsomedial aspect of MC3 that contained the most dense bone (Young et al. 2007). These findings correlate with the area of contact we found under impact loading in our study and with the medial side withstanding a larger total area of loading when compared to the lateral condyle. It is possible that the remodelling that is occurring within the distal end of MC3 could be the result of not only midstance loading, but additionally the impact trauma that occurs during primary impact. It has been suggested that high-frequency vibrations orchestrate a bone response that produces a stiffer, thicker trabecular structure (Ozcivici et al. 2007), similar to the frequencies measured previously in the equine distal limb under simulated impact loading (McCarty et al. 2014).

3.6.2 Contact Area

Under impact loading, with a MCP joint angle between 165-175°, we found the contact area on the condyles on MC3 being loaded by P1 to be almost three times larger than the loading from the proximal sesamoids. This finding contrasts with investigations of midstance which show P1 contact to be less than twice PS contact (Easton et al. 2007, Brama et al. 2001). This difference is largely due to the limb orientation occurring at impact compared to that at midstance. Easton et al. 2007 found the total contact area at midstance from P1 and the proximal sesamoids to be between 67% and 87% of the total joint surface, depending on speed, with the proximal sesamoids and intersesamoidean ligament accounting for 42-46% of the contact on the condyles of MC3 depending on joint angle. As the speed increased, the proximal
sesamoids accounted for more of the contact area on the condyles of MC3 therefore absorbing more force over a smaller area. These results were similar to those measured by Brama et al. 2001 who found that at 1800 N (stance) approximately 63% of P1 was in contact with the distal condyle compared to a gallop (10,500N) when 87% of P1 made contact with MC3.

3.6.3 Impact Loading and Joint Degradation

Subchondral bone sclerosis within the distal end of the equine MC3 is common among racehorses and is thought to be representative of the loading history. The focus of previous research has midstance loading to be primarily responsible for the bone adaptation that occurs in the distal end of MC3 (Easton et al. 2007, Brama et al. 2001, Colahan et al. 1988). Our results indicate that loading at impact should also be considered, especially as high-frequency, high-magnitude impact loading has been shown to elicit changes to the bone micro-architecture (Radin et al. 1972, 1984, Garman et al. 2007). Given the combination of high energy loading over a small area of contact, it is possible that impact loading may contribute to the remodelling and sclerosis of the subchondral bone within the MCP joint. Impact loading has been shown to cause an increase in bone stiffness (Radin et al. 1972), reducing the ability to absorb subsequent loading energy and contributing to overall joint degradation over time (Young et al. 2007).

In a study conducted simultaneously with the current study (McCarty et al. 2014), we found hoof strike to have a significant effect on the accelerations and frequencies measured from hoof to MC3. Although hoof strike did not significantly affect the joint contact area, the manner in which a horse makes contact with the ground could have an effect on the predisposition of the bone to damage due to the high magnitude shock waves being transferred through a small contact area under impact loading.
3.6.4 Limitations

Failure to preload the flexor and extensor tendons in the distal forelimb may have created a source of error within the experimental set-up. It has been shown that although tension in the digital flexors and extensors does exist prior to hoof colliding with the ground, there is almost no rotation of the MCP joint at that time suggesting that the forces applied are equivalent but opposing (Harrison 2012). Although hoof strike upon primary impact did not significantly affect the measures of this study, it may be of importance to soft tissue injury.

The issue of the film’s range of sensitivity was discussed in the methods. Since the joint was mildly disrupted in order to insert the film into the joint, this could also have introduced some error into the measurements taken.

3.7 Conclusion

This is the first study that has examined the contact area within the MCP joint under impact loading. Under impact, contact area and joint loading primarily occurred between P1 and MC3 compared to midstance loading where the joint is equally loaded by P1 and PS (Figure 3.8.5). Although manner of hoof strike did not have a significant effect on contact area, previous work has shown that strike does affect acceleration magnitude and frequency. Based on these results, it is possible that impact could have a role in the etiology of OA and should be considered for future study. Quantification of contact pressure magnitudes and stress distribution patterns within the MCP joint under impact loading would be recommended for future study.
Figure 3.8.1 - Experimental apparatus used to simulate primary impact. Dark line on hammer head face represents the rubberized contact surface.
Figure 3.8.2 – Distal aspect of a third metacarpal. Blue areas indicate placement of pressure sensitive film within the MCP joint including P1 and PS contact area. Light blue area represents lateromedial film used to capture loading across the sagittal ridge.
Figure 3.8.3 - Sample film that has been thresholded according to a calibration and segmented into pressure levels, with corresponding greyscale intensities. The over limit level was used for data analysis as it provided defined boundaries (as indicated by the steep drop off within the greyscale intensity) that did not extend beyond the film size and were therefore confident in capturing a representative contact area.
Figure 3.8.4 – Global mean (red/blue dot) and range (surrounding ellipses) for $C_x$ and $C_y$ under all strike conditions (Table 2) in both the lateral and medial condyles for areas loaded by P1 and the proximal sesamoids (PS).
Figure 3.8.5 – A comparison of A) the joint contact area during midstance loading under an ex vivo simulated galloping load of 10500 N on P1 (Brama et al. 2001) and B) the contact area found in the current study during impact loading on P1 represented by the lighter shading. C) A comparison of contact area at midstance at a MCP joint angle of 120-150 degrees on MC3 using a dye staining technique (Easton et al. 2007) and D) the contact area found in the current study under impact loading as indicated by the red area on the distal end of MC3. Overall the contact area in midstance loading is much greater and distributes across the palmar aspect of P1 and MC3 compared to the contact area for impact loading.
**Figure 3.8.6** - Comparison of P1 contact area according to hoof strike. While the trend line indicates an increase in total contact area from a heel to flat to toe strike, none of the pairwise comparisons among them were statistically significant (p-values 0.58, 0.54, 0.69).
Table 3.1 – The median and lower/upper limits of the over-limit joint contact area by location (lateral/medial condyle), hoof strike (heel/flat/toe strike) and bone (P1, PS).

<table>
<thead>
<tr>
<th></th>
<th>LL area (cm²)</th>
<th>Median area (cm²)</th>
<th>UL area (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>lateral condyle</td>
<td>0.70</td>
<td>1.09</td>
<td>1.70</td>
</tr>
<tr>
<td>medial condyle</td>
<td>0.76</td>
<td>1.19</td>
<td>1.85</td>
</tr>
<tr>
<td>heel strike</td>
<td>0.62</td>
<td>0.97</td>
<td>1.52</td>
</tr>
<tr>
<td>flat strike</td>
<td>0.77</td>
<td>1.22</td>
<td>1.94</td>
</tr>
<tr>
<td>toe strike</td>
<td>0.75</td>
<td>1.24</td>
<td>2.05</td>
</tr>
<tr>
<td>P1</td>
<td>1.23</td>
<td>1.91</td>
<td>2.97</td>
</tr>
<tr>
<td>PS</td>
<td>0.43</td>
<td>0.68</td>
<td>1.06</td>
</tr>
</tbody>
</table>

Table 3.2 (global means) - The distances between the centroids and the sagittal ridge ($C_x$, cm) and the transverse ridge ($C_y$, cm) on each condyle and the dispersion of pixels within each area relative to the centroid ($I_{xy,n}$, cm⁴).
CHAPTER 4: A Comparison of Stress Distribution in the Healthy and Diseased Equine Third Metacarpal under Impact and Static Loading Using Finite Element Analysis

Authors and Contributions:

Cristin A. McCarty – The University of Guelph (Department of Biomedical Science) The main contributor to study design, performed the majority of the study execution, the majority of the data analysis and interpretation, and main contributor to the manuscript preparation with suggested revisions from the other authors.

Jaques Milner- The University of Western Ontario (Robarts Research Institute) Made a significant contribution to the study design, a significant contribution to the study execution (materials algorithm, guidance with Abaqus) a significant contribution to data interpretation and provided guidance on manuscript preparation.

Karen Gordon - The University of Guelph (School of Engineering, Department of Biomedical Engineering) Made a significant contribution to the study design, minimal contribution to the study execution (use of Abaqus), a minimal contribution to data analysis and interpretation and provided guidance on manuscript preparation.

Jeffrey J. Thomason - The University of Guelph (Department of Biomedical Science) Made a significant contribution to the study design, a significant contribution to the interpretation and provided guidance on manuscript preparation.

Timothy Burkhart - The University of Western Ontario (School of Engineering, Department of Biomedical Engineering) Made a significant contribution to study design, a minimal contribution to study execution and provided guidance on manuscript preparation.

David Holdsworth - The University of Western Ontario (Robarts Research Institute) Made a significant contribution to the study execution (μCT scanner), a minimal contribution to data analysis and interpretation and provided guidance on manuscript preparation.
**4.1 Abbreviations**

MC3 – Third metacarpal

FE – Finite element

OA – Osteoarthritis

MCP – Metacarpophalangeal

µCT - Micro computed tomography

PSG – Parasagittal groove

P1 – First Phalanx

PS – Proximal sesamoids

SR – Sagittal ridge

MPa – Megapascals

STL – standard triangulation language

3D – Three dimensional

E – Elastic modulus

ρ - Apparent bone density
4.2 Abstract

**Reasons for performing study:** To determine whether stresses induced by impact were comparable with those under static loading in healthy and diseased bone, to assess the potential role for injury based on the stress magnitude and location.

**Objectives:** To compare the stress distribution in the distal third metacarpal (MC3) in two specimen specific finite element (FE) models (1 healthy/1 osteoarthritic (OA)) of the equine metacarpophalangeal (MCP) when loaded under static and impact conditions.

**Study Design:** Two 10-node quadratic tetrahedral FE models of the equine MCP joint (healthy/OA) were generated from micro-computed tomography (µCT) images and loaded under simulated midstance (static) and primary impact (impact) conditions.

**Methods:** Both models were built and loaded using FE software. Each model was loaded statically and under impact conditions. Results were validated using previous data from ex vivo experiments. Comparisons were made between stress magnitudes and distributions between models and loading conditions. Data were analyzed from a standardized palmar and dorsal slice of the distal end of MC3, which was further divided by location: lateral condyle, lateral parasagittal groove (PSG), sagittal ridge (SR), medial PSG and medial condyle. Von Mises stresses (MPa) and contact pressures (MPa) were reported where possible.

**Results:** Under static loading, the highest stresses – maximum (32.0 MPa) and average (19.38 MPa) - were located in the palmar PSG of the healthy model. Under impact, the OA model had the highest stress across all locations, with the greatest stress by location on the dorsal aspect of the medial condyle (14.1MPa).
**Conclusions:** Overall, the highest stresses were found on the palmar aspect of MC3 in the PSG of the healthy model under static loading. In general, loading on the dorsal aspect of MC3 had the highest stresses associated with impact loading in the OA model. The stress magnitude and distribution with the distal end of MC3 is dependent on both the health status of the bone as well as the loading being applied.

**Potential relevance:** Our results suggest that impact loading should be considered in the context of injury as a result of mechanical overload.

**4.3 Introduction**

The metacarpophalangeal (MCP) joint of horses is a valuable large-animal model for studying bone disease of mechanical origin. Osteoarthritis (OA) in the joint is common among racehorses (Parkin et al. 2006, Neundorf et al. 2010) and is associated with change in the micro-architecture of the subchondral bone and overall joint geometry (Easton 2012, Muir et al. 2008, Drum et al. 2007). Exercise-induced bone remodeling in the condylar region of the third metacarpal (MC3) can cause subchondral bone sclerosis and damage to the articular cartilage, leading to degenerative lesions common to advanced OA among mammalian species (Radin et al. 1972). These lesions weaken the joint integrity and can lead to catastrophic condylar failure that ultimately results in early retirement or euthanasia of young horses (Parkin et al. 2006).

The underlying bone structure within the MCP joint is representative of the mechanical loading history (dependent on magnitude, rate and repetitiveness) sustained during high-speed racing and training (Young et al. 2007, Norrdin et al. 1998). The unique joint configuration – high range of motion and small surface area – involves multiple loading sites as the joint moves.
from flexion into extension while under load. Contact stress during midstance (Figure 4.8.1) (i.e., at full joint extension) has been shown to be associated with site-specific changes within the distal end of MC3 (Young et al. 2007, Riggs et al. 1999). Previous modelling of the mechanics of the MCP joint shows that the distal condyle of MC3 is pinched between the first phalanx (P1) and proximal sesamoid bones (PS) at midstance (Figure 4.8.1 – inset), creating a combination of high compressive and shear loading in the dorsal and palmar aspects of MC3 (Easton 2012).

Force magnitudes are considerably lower at impact (approx. 2-10% of the peak at midstance) (Figure 4.8.1), but it is known that repetitive impact can be involved in the etiology of osteoarthritis (Radin et al. 1972, 1984, Gustås et al. 2006, Serink et al. 1977). Accelerations of high magnitude and high frequency have also been shown to elicit bone changes and contribute to damage within a joint (Radin et al. 1972). Repetitive impact loading under high speed locomotion occurs 100-150 times per minute during racing and training, and imposes large stresses on the MCP joint. Over time, stresses such as this are known to cause microcracks in the SBC and calcified cartilage initiate remodeling which leads to stiffening of the subchondral bone, endochondral ossification, tidemark advancement, resulting in thinning of the articular cartilage and increased stresses at the base of the articular cartilage which eventually causes damage to the overlying cartilage (Burr and Radin 2003).

Impact and static loading at the various phases of the stance in the horse has been well documented (Gustås et al. 2006); stresses and strains have been presented for the midshaft of MC3 at midstance (Biewener et al. 1983, Nunamaker et al. 1990), and contact pressures and midstance stresses in bones of the MCP joint have been assessed (Easton et al. 2007, Brama et al. 2001, McCarty et al. 2014b). This study builds on these previous in vivo and ex vivo studies. Its purpose is to create two subject-specific, three-dimensional, finite-element (FE) models of the
equine MCP joint (one with advanced OA and one healthy) and compare the stress distribution patterns under impact and static (midstance) loading in each model.

4.4 Materials and Methods

4.4.1 Image acquisition and segmentation

Images were acquired from a micro-computed tomography (µCT) scan of two right equine MCP joints (one healthy and one exhibiting signs of advanced OA – bone sclerosis, erosion of the articular cartilage and deep pitting in the palmar aspect of the subchondral bone on MC3) from female Standardbred horses (ages 5 and 7) that were euthanized for reasons unrelated to the musculoskeletal system and were harvested post mortem for use in this study. The scans consisted of the third metacarpal (MC3) and the proximal phalanx (P1), but not including the proximal sesamoids (PS). Each limb was oriented so that the MCP joint angle was between 165-175°, similar to that found at primary impact in the live horse (Back et al. 1995, Clayton et al. 1990, Linford et al. 1994). The scan was performed using a µCT scanner (GE Locus Ultra, GE Healthcare, Milwaukee, WI) located in the Robarts Research Institute at the University of Western Ontario in London, Ontario, Canada. Data were collected with 120kV, 20mA and 900 views, generating isometric voxels of 0.154mm. The reconstruction voxel size was 0.154mm³. The images were imported into Amira 5.2.2 (Mercury Computing Systems, Chelmsford, MA) where a 3-dimensional (3D) surface of the bone was reconstructed, smoothed and exported as a standard triangulation language (STL) file for importing into the FE software.

4.4.2 Model features

FEM presents a challenge to the user in creating a model that provides sufficient detail (which can introduce errors and increase computation time) and over-simplifying the model (through assumptions made when specifying parameters of the model). It is important to keep in
mind the goal of creating the model and the purpose of the comparisons. While there have been
highly detailed FE models of the human foot capable of simulating loading conditions under
various phases of the stance of the human foot while walking (Quin et al. 2013), such models
lack specimen specificity in the micro-architecture of structures. Likewise, specimen-specific
FE models may provide less structural detail, including the omission of associated structures
(ligaments, cartilage) within a joint, while having detailed material properties (e.g.
inhomogeneity). The level of detail within each model is representative of the purpose of the
experiment. It is impractical to build a FE model that includes all of the associated structures
and the individual detail due to the high computational cost and would generally be considered
unnecessary given the question being asked. For the purposes of this study, the models include
P1 and MC3. Load transfer from the proximal sesamoids is simulated in the static loading
condition, and is not necessary for the impact loading condition, based on our previous ex vivo
testing of the contact area within the MCP joint at impact (McCarty et al. 2014b). The use of
spring elements to model tendons and ligaments tend to oversimplify the model and introduce
error by neglecting to account for tendon or ligament structure and loading rate. The lack of PS
contact with MC3 (as indicated by the pressure film results in chapter 3) indicates that the
tendons associated with the MCP joint were not active during this phase due to MCP joint
orientation at primary impact. Although it has been shown that the flexor and extensor tendons
in the equine distal limb maintain some tension in order to properly align the joint at impact,
since there is no joint rotation at this moment (Harrison et al. 2012), the net tendon forces were
assumed to be zero. Furthermore, since primary impact occurs more quickly (3-5 ms) than the
neuromuscular response time (50ms), the tendons were considered to be acting passively at this
phase of the stance (Lanovaz et al. 1998, Thomason et al. 2008), and were not included in the
model. Under high-rate loading, articular cartilage compressive stiffness can increase up to an order of magnitude (Jeffrey et al. 1995) and it is possible that it could stiffen to a magnitude similar to that found in SCB (personal communication with Micheal Buschmann). For this reason, the articular cartilage was not included in the model.

4.4.3 Mesh generation

Each 3D surface was imported into an automated mesh generating software program (NetGen, Linz, Austria) where it was converted to a 3D volume mesh that consisted of 4-node linear tetrahedral elements (Table 4.1). These elements were chosen because they can be generated through an automated process (unlike hexahedral meshes which require manual processing) and provided a mesh density sufficient to capture the heterogeneous material properties present in both MC3 and P1. The meshed surfaces were then imported into a finite element analysis software package (Abaqus 6.12, Simulia, Providence, RI). Once the model had been imported into Abaqus, the elements were converted to modified 10-node quadratic tetrahedral elements, to reduce the stiffness often associated with 4-node elements, while producing accurate results for compact and cancellous bone (Ramos and Simoes 2006).

4.4.4 Material properties

The inhomogeneous bone material properties were determined by mapping each voxel using custom written software and user-defined parameters (Austman et al. 2008). This algorithm, \( E = a + b \rho^c \) (where \( E \) is the elastic modulus (MPa) and \( \rho \) is the apparent bone density (g/cm\(^3\) or g/ml)) was developed in a previous study (Austman et al. 2008), and was chosen for use here based on its ability to accurately predict material properties in a long bone through a comparison of FE strain using multiple density-modulus equations and experimental strain. The algorithm uses the well-known relationship between Hounsfield intensity values—indicating
bone density in μCT scans—and orthotropic elastic moduli that have been established for the metaphyseal and subchondral bone in the distal condyles of MC3 (Rubio-Martinez et al. 2008, Les et al. 1994). The equation used in this study was: $E = 9040\rho^{2.35}$, with coefficients well within the range of previously developed density-modulus equations. It also extrapolates well to the maximum apparent density for equine specimens: 2.47 g/cm$^3$ compared to human bone which has a maximum of approximately 2.0 g/cm$^3$ (Figure 4.8.5). The equation essentially performs a four-stage conversion—from image intensity, to ash density, to apparent density, to elastic modulus and Poisson’s ratio—and provides the necessary material properties to create a subject specific finite element model (Figure 4.8.2).

Sensitivity testing was done to determine the effect of bone stiffening that occurs under an increased loading rate. The algorithm used to map the material properties (Figure 4.8.5) was multiplied by a factor of 1.5, based on the relationship known between strain rate and Young’s modulus in trabecular bone (Linde et al. 1991). This test was performed in order to account for the stiffening of the bone with an increase in strain rate that occurs under impact loading. Although there were slight changes to the model on an element-by-element basis, there was no distinguishable change to the average stress within a given location.

A set of very dense elements were specified (density = $5.0 \times 10^{-7}$, Young’s modulus = 16000 MPa, Poisson’s ratio = 0.3) to the proximal cut end of MC3 in order to increase the overall mass of MC3 to 5kg. This increased the mass of the model to represent the effective mass associated with impact loading in the live horse (Thomason et al. 2008).

**4.4.5 Loading and Boundary Conditions**

Each model was loaded under both a static and an impact load. Static loading was created using Abaqus Standard software, and was applied using a pressure load to a specified
number of nodes on the distal aspect of MC3 representing the areas where P1 and PS articulate under midstance loading (Table 4.2). The orientation of the loading followed the curvature of the bone so the loading was always applied at 90° to the bone surface. The pressures and locations were chosen based on ex vivo data collected in the MCP joint for contact pressures and areas under midstance loading (Easton et al. 2007, Easton 2012). The proximal cut end of MC3 was constrained in all directions.

Impact loading was created using Abaqus Explicit software, by applying a velocity to all nodes on MC3, effectively asking the software to collide this bone with a stationary P1 (the distal end of which was constrained in all directions). Positive x was directed laterally (velocity = 0 m/s), positive y was directed palmarly (velocity = 0 m/s) and positive z was directed distally along MC3 (velocity = 3.55 m/s). Contact was defined as general surface-to-surface contact with a coefficient of friction between P1 and MC3 set at 0.007 (Nobel et al. 2011). A linear interaction property was used with the stiffness surface properties defined at 12 MPa/mm to allow for sufficient contact and settling of the contact surfaces before separation was allowed for between the two bones. The total impact duration was set at 3 ms based on live animal data collected under high speed movement (Thomason et al. 2008).

4.4.6 Convergence

FE analysis can take hours to run a simulation and is sensitive to many of the conditions defined by the user. It is therefore important to converge the model in order to ensure accurate predictions and minimize the overall computational time to run a simulation. Convergence occurs when a low mesh density (coarse) is compared to a higher mesh density (fine) and the results from both models once loaded is within the convergence criterion (Logan 1986). A convergence analysis was performed to determine the optimal mesh density in which,
coarse (low density), moderate (medium density) and fine (high density) meshes were generated. Sensitivity to resolution of specimen specific bone material properties, comparison of contact area and von Mises stress at multiple areas on the distal end of MC3 were used to determine the appropriate mesh density for the FEM used in this experiment. Convergence criterion was defined as ±5% the difference from the highest resolution mesh.

4.4.7 Validation

The impact model was validated using data from contact pressures and locations obtained from ex vivo testing. A detailed description of the experimental apparatus and protocol can be found in McCarty et al. (2014b) however it should be noted that for the purposes of this experiment, the data from the pressure films used here was obtained from a separate testing where a higher range of pressure film (2.5 – 10 MPa) was used. Briefly, pressure sensitive film was placed within the MCP joint of equine cadaver forelimbs. Contact pressure and contact area between P1 and MC3 were estimated by the film under simulated impact loading from a 24 kg pendulum impact hammer. The height at which the hammer was released produced a repeatable impact velocity of 3.55 m/s at contact, which is within the normal in vivo range: 1.43 m/s for a medium trot to 7.2 m/s for a racing trot (Wilson et al. 2001, Lanovaz et al. 1998, Gustás et al. 2001). The static model was validated using contact pressures that were taken from ex vivo data in the literature where simulated mistance loading was applied to the distal equine limb and contact pressures were recorded for a given area (Easton et al. 2007, Easton 2012, Brama et al. 2001).

4.4.8 Data Analysis
The distal end of MC3 was subdivided into eight anatomical locations (Figure 4.8.2) for the purposes of comparison between the OA and healthy models and between static and impact loading. Two mediolateral slices were extracted from a standardized location on the palmar and dorsal aspect on the distal end of MC3 in each model and loading condition. A map of 160 equidistant points was superimposed over each mediolateral slice and the von Mises stress was recorded at each point (Appendix B). Images of each slice were standardized to equal size, so that when the point map was superimposed, the location was comparable between models and loading conditions. The point map was also used to map the material stiffness at each point of the slices, to compare with the resulting von Mises stress.

4.5 RESULTS

4.5.1 Convergence

Three models ranging from 33 905 to 1 603 8345 elements were created (Table 4.3). All three models were identical in the element type, the loading and the boundary conditions applied to each model, and differed only in the mesh density and relative material property distribution (Figure 4.8.4). The results show the comparison of the von Mises stress distribution at impact between the lower resolution models and the highest resolution model (Figure 4.8.4). The results of the convergence analysis, using the von Mises stresses (Figure 4.8.4) determined that the moderate resolution model obtained convergence within ±5% based on the average stress within a given location from a slice on the dorsal aspect of MC3 (Table 4.4). Based on these results, the moderate mesh was determined to produce a model with sufficient resolution of the material properties based on the micro-architecture (Figure 4.8.3) and the von Mises stress distributions that was within the convergence criterion (Table 4.4).
4.5.2 Validation

Static loading was validated by comparing the current models (healthy and OA) to Easton’s (2012) models (CTL – control model that represented a healthy bone and NFX – a non-fractured model with OA) (Figure 4.8.6a and 4.8.6b). While the geometry of our models differed from those of Easton 2012, our results show very similar stress distribution and magnitudes between the healthy model and Easton’s CTL model, as well as between our OA model and the NFX model (Figure 4.8.6a). The surface von Mises stress showed a similar distribution to those found by Easton 2012, with the area of highest surface stress in the palmar aspect of the distal MC3 where the PS make contact. Furthermore, when we compare the average von Mises stresses at a 30° palmar location, within anatomical areas associated with PS contact (Figure 4.8.2B), our results are very similar to those found by Easton 2012 under static loading (Figure 4.8.6b). Although the stress magnitudes were not identical between the current models and Easton’s models, Easton reported that the FE model tended to overestimate contact stresses when compared to the ex vivo data. Overall our models had good agreement of von Mises stress within location to Easton’s models with the exception of the lateral parasagittal groove which we found to be greater in both the healthy and OA models (Figure 4.8.6b).

Impact loading results were validated using experimental ex vivo data. Despite our best efforts to capture the maximum joint pressures in the MCP joint under impact loading using the higher range of film, our results found that the pressures were over-limit of the range of the pressure film. The results from the impact models and testing between both FE models and the experimental ex vivo testing showed similar contact locations with a defined border of contact ending at the transverse ridge (Figure 4.8.7a). While there was little difference between average contact pressure between the healthy and OA FE models at each location, both models tended to
overestimate the contact pressure when compared to the experimental results (Figure 4.8.7b). This difference may not be as extreme as the figures indicate, because an accurate maximum pressure from the PF in the experimental testing was unobtainable: the pressure range of the film was exceeded over most of the contact area.

4.5.3 Static Loading Stress Distribution

The highest von Mises stresses (MPa) under static loading were found within the palmar PSG area that is associated with loading from PS, for both healthy (range 9.688 – 32.0 MPa) and OA (range 6.429 – 25.6 MPa) models (Figure 4.8.6a and 4.8.6b). The palmar aspect of the healthy model had higher average stresses (19.38 MPa and 19.34 MPa) at the lateral and medial PSG locations, compared to the OA model (14.34 MPa and 12.24 MPa), at the same location (Figure 4.8.6a and 4.8.6b). This indicates greater stress in the healthy model even when the total loading area was controlled for between models (Table 4.2).

4.5.4 Impact Loading Contact Pressure and Contact Area

The average contact pressure between the healthy and OA model were found to be very similar across locations with the highest pressures occurring in the lateral PSG (11.2 MPa; Figure 4.8.7a and 4.8.7b). This was similar to the findings in the experimental ex vivo testing, which showed peaks at 7.25 MPa (Figure 4.8.7b). Maximum contact pressure (25.46 MPa) for the healthy model was located in the medial PSG and medial condyle and in the lateral and medial PSG of the OA model (33.01 MPa; Figure 4.8.7b).

The contact area in the FE models was similar to the results found in the experimental ex vivo testing (McCarty et al. 2014b) with well-defined borders of contact from P1 up to the sagittal and transverse ridges and contact occurring across the sagittal ridge (Figure 4.8.7a).

4.5.5 Impact Loading Stress Distribution

83
Surface von Mises stress patterns were similar to the contact stress distribution in the healthy and OA models (Figure 4.8.8a). The average von Mises stress across all locations, in both the palmar and dorsal slice was found to be greater in the OA model when compared to the healthy model under impact loading (Figure 4.8.8b). The highest stresses were located in the medial condyle at 12.8 MPa and 14.1MPa in the healthy and OA models respectively, while the sagittal ridge was the location associated with the lowest stresses at 6.1MPa and 8.2MPa in the healthy and OA models respectively (Figure 4.8.8b).

**4.5.6 Static vs. Impact Loading Contact Pressure**

We were unable to obtain contact pressures from the statically loaded models in the current study given the model construction and the method used for loading. A comparison of the impact contact pressures to the statically loaded FE models and experimental models previously reported (Easton 2012) indicate that the contact pressures on the medial and lateral condyles under impact are similar to those found at static loading during a gallop (Figure 4.8.9).

**4.5.7 Static vs. Impact Loading Stress Distribution**

Within the palmar side of distal MC3, the highest von Mises stresses were associated with static loading in the healthy model at all locations, while the lowest stresses occurred in the healthy model under impact loading (Figure 4.8.10a). The highest stress by location within the palmar aspect of MC3 was the lateral PSG with an average von Mises stress of 19.9MPa (Figure 4.8.10b). The loading on the dorsal aspect (associated with contact from P1), found the highest von Mises stresses occurred under impact loading in the OA model across all locations with the exception of the lateral condyle in which static loading in the healthy model produced the highest stress at 14.5MPa (Figure 4.8.10c).
4.6 DISCUSSION

This study used specimen specific models to compare the stress distribution within the distal end of MC3 under static and impact loading. Although midstance (static) loading has been the focus of previous studies that have examined stress distribution under simulated loading using FE in the equine forelimb (Les et al. 1997, Hinterhofer et al. 1997/2001, McClinchey et al. 2003, Collins et al. 2009, O’Hara et al. 2012 and Easton 2012), this is the first study to examine impact loading within the equine MCP joint. Results from this study support the hypothesis that each FE model, regardless of health status, produces noticeably different stress patterns within the distal end of MC3 when loaded under impact and static conditions. Health status (healthy or osteoarthritic) had an effect on stress magnitude and distribution depending on the type of loading (static or impact).

4.6.1 Limitations

When using FE modeling many assumptions are inherently made in building, constraining and loading the model. Simplifications are necessary when modeling biological systems due to the complexity in the materials and structures involved. Although there are many other structures present in the equine MCP joint (articular cartilage, synovial fluid, ligaments, etc.), a model that includes all of the structures would have a high computational cost, and such detail is not entirely necessary because the primary forces on MC3 are across the articular surfaces. For this reason, only the bone structures involved in impact loading (P1 and MC3) and specimen specific material properties were included in the model. While we were able to apply material properties based on the bone density, bone material was modeled as being isotropic due to the constraints imposed by the Abaqus Explicit solver.
Although care was taken to create a displacement boundary condition that was well away from our area of interest, some of the resulting high stresses found on the proximal end of MC3 were likely due to the constraint occurring from the boundary condition rather than the contact and loading stresses. These were away from the regions of interest at the distal articular surface.

Validation is an important factor in determining the accuracy of a FE model in order to have meaningful results. The contact area and distribution from our ex vivo results were very similar to our results in the present study, although we were unable to obtain maximum MCP joint contact pressures under ex vivo impact loading. Further work is needed to accurately assess the maximum joint contact pressures occurring in a live animal through ex vivo testing under varying impact velocities.

4.6.2 Static Loading Stress Distribution

The pinching action that occurs under midstance loading between P1 and PS on the distal end of MC3 is a combination of compressive and shear loading that creates high stresses within a localized area (Easton 2012). Previous authors have suggested that this stress localization that occurs under midstance loading is responsible for the onset on osteoarthritis within the MCP joint of performance horses (Kawcak and McIlwraith 1994; Colahan et al. 1988). Our results showed similar stress patterns to those found by Easton (2012) with higher stresses found in the palmar aspect of MC3, the area normally loaded by the PS during midstance. The higher stresses associated with the palmar aspect of MC3 compared to the dorsal region are likely the effect of applying greater pressure to the area loaded by PS compared to P1 based on the results from ex vivo testing (Easton 2012). A comparison by location found that the highest stresses were found in the PSGs compared to Easton (2012), who found the highest stresses to be in the condylar
region of MC3. This difference may be due to the individual specimen bone geometry and the
difference in user specified parameters used to build the model. Easton (2012) modelled the
contact between P1, PS and MC3 and applied a concentrated force load to the proximal end of
the model. The loading conditions in the current study used a pressure load that was applied to
the distal end of MC3 given the contact pressure and location of PS and P1 based on
experimental results (Easton 2012). The high stress regions located in the PSG of the healthy
model of the current study coincide with a region of low bone density and could be the result of a
bone density gradient at this location. It has been well documented that the condyles become
more dense than that at the SR, which could lead to a gradient which may cause an increase in
the stress and strain at the interface between the condyles and the SR (Riggs et al. 1999, Rubio-
Martinez et al. 2008).

The overall bone density and material stiffness of the healthy bone within the condylar
region was found to be less than in the OA bone (Figure 4.8.3) and therefore would likely allow
for greater strain per unit area due to the decrease in bone stiffness. Under normal loading
conditions, subchondral bone acts as a shock absorber and is able to readily deform in order to
absorb much of the energy being transferred within a joint while incurring some micro-damage
to the internal structure of the bone. This microdamage initiates bone remodelling in order to
repair the bone damage and increase the strength of the trabecular bone to an attempt to resist
fracture. The result is bone stiffening, making it less able to readily deform therefore decreasing
its ability to strain and absorb subsequent loading energy.

4.6.3 Contact Pressure and Contact Area
There was no effect on contact pressure when comparing the healthy and OA models under impact loading. This is to be expected, as much of the differences in bone stiffness occurred beneath the bone surface within the underlying subchondral bone (Figure 4.8.3). Contact pressures between the impact in our study and those determined under static loading (Easton 2012) were found to be similar (Figure 4.8.9). The models loaded under impact were given a loading rate of 3.55m/s, which would be similar to a medium trot in the live animal. This velocity was chosen based on the experimental velocity applied to the ex vivo testing that was used for validation in the current study, however it is likely that if these models were loaded using a galloping or racing trot impact velocity that the values would exceed the contact pressures determined by Easton (2012) under midstance loading. The FE models in the current study were found to overestimate the contact pressures when compared to the experimental data, however this difference is difficult to quantify as the experiment pressure film was unable to identify pressured beyond 10MPa due to a limitation in the range of the film. Similar to our results, Easton’s (2012) models tended to overestimated values compared to the experimental data by up to 50% (Medial and lateral P1 and medial PS at gallop loads). This could be associated to a variety of factors used in building the FE models including but not limited to the element choice, the density-modulus algorithm chosen or lack of structures present in the MCP joint (ie. synovial fluid).

The contact area associated with impact loading was found to occur primarily between P1 and MC3 compared to midstance loading where both P1 and PS apply contact pressure on MC3. The orientation of the distal limb at primary impact allows for a MCP joint angle of approximately 165-175° based on in vivo kinematic data (Back et al. 1995, Clayton et al. 1990, Linford et al. 1994). As was determined experimentally (McCarty et al. 2014b), the PS do not
make significant contact with MC3 during this phase of the stance. As our results indicate, the dorsal aspect of MC3 is primarily loaded during impact (Figure 4.8.7a) compared to the contact area on both the palmar and the dorsal aspect of MC3 under midstance loading.

4.6.4 Impact Loading Stress Distribution

Impact loading in the distal equine limb has been shown to produce high-magnitude, high-frequency signals that contain significant energy extending up into the mid-diaphysis of MC3 (McCarty et al. 2014a). It has been suggested that high-frequency vibrations orchestrate a bone response that produces a stiffer trabecular structure (Ocvivici et al. 2007) initiating a cascade of events that eventually lead to the thinning of the articular cartilage and increased risk of shear stress to the overlying cartilage (Burr and Radin 2003). In the current study, impact loading was found to produce higher stresses in the OA model when compared to the healthy model likely due to the increase in bone stiffness associated with a horse diagnosed with OA. Analysis of subchondral bone within the MCP joint of racehorses has shown increased bone density in certain areas of MC3 causing an increase in bone stiffness in response to mechanical loading (Rubio-Martinez et al. 2008, Drum et al. 2008, Muir et al. 2008). It is possible that horses with OA may be at greater risk to overload bone and cartilage injuries due to the increased bone stiffness and associated higher stresses observed under impact loading.

4.6.5 Static vs. Impact Loading Stress Distribution

The yield stress in the lateral condyle, medial condyle and sagittal ridge has been reported as 113.3 +/- 3.115, 116.2 +/- 3.105 and 65.75 +/- 3.136 MPa respectively in the distopalmar aspect of the equine MC3 (Rubio-Martinez et al. 2008). The von Mises stress in the
current study were all found to be well below the yield stress at any location and therefore not likely to cause ultimate bone failure under compressive loading.

Bone responds to mechanical stimuli by remodeling to repair damaged bone and to strengthen the areas in the direction that the primary load is applied (Frost et al. 1990, Easton et al. 2008). It has been shown that areas consistently in contact under higher loads under midstance loading are associated with increased subchondral bone density suggesting that SCB remolds and adapts to the applied load (Easton et al. 2007). An in depth analysis of the distal MC3 of horses with OA showed that the dorsomedial aspect of MC3 contained the most dense bone (Young et al. 2007), which coincides with our results from both models under impact loading with the dorsomedial condyle undergoing the highest stresses when loaded.

Although the highest stresses occurred in the OA model under impact loading in most locations, the highest stress was associated with the lateral condyle in the healthy model under static loading. It is likely that the changes that occur in the palmar aspect of MC3 are due to the high stresses produced by the PS during midstance (static) loading due to the lack of loading that occurs in the palmar aspect of MC3 under impact loading (McCarty et al. 2014b). Loading on the dorsal aspect of MC3 occurs during impact and midstance (static) loading and creates similar stress magnitudes under each loading condition. While the lowest stresses were found to be associated with impact loading in the healthy model on the dorsal aspect of MC3, the increased bone stiffness in the OA model, created stresses under impact loading that exceed those found in static loading. This suggests that although it is unknown which modality of loading may lead to the increase in bone stiffness, once these changes occur, impact loading may induce further damage to the underlying SCB and the overlying articular cartilage. Although impact loading has been shown to increase bone stiffness (Radin et al. 1972), in the healthy bone of the current
study, the higher stresses occurred under static loading indicating that the bone may undergo remodeling in order to strength the SCB structure and resist damage due to mechanical overload. While the remodeling and increase in bone stiffness appears to be adaptive in response to static loading – as indicated by the reduction in stress in the OA model – under impact loading, the increase in bone stiffness reduces the ability of the SCB to absorb subsequent energy due to the impaired viscoelastic response under high-rate loading.

4.7 CONCLUSION

To the best of the authors’ knowledge this is the first study to examine the effect of impact loading on the MCP joint in horses using FE analysis. Both the loading that occurs during impact and midstance are likely to contribute to the cascade of events that overtime lead to OA within the MCP joint. There are many factors that play a role in biomechanical loading and joint injury including individual conformation, footing surface, neuromuscular fatigue and speed and duration of training and racing (Thomason et al. 2008, Cruz et al. 2008). The role of impact loading on joint mechanics has proven to be significant and worthy of study in the future, however the role of impact on the etiology of OA within the MCP joint of performance horses requires further examination and cannot be commented on based on these findings.
4.8 Figures

**Figure 4.8.1** – Phases of the stance and associated loading conditions at each phase. Red arrows indicate vertical and horizontal acceleration and blue arrows indicate the ground reaction forces associated at each phase of the stance. The length of the arrow represents the magnitude of the acceleration and/or force at a given phase. Inset below midstance – green arrows indicate pinching action that occurs on MC3 from P1 and PS. Figure modified from Thomason and Peterson 2008.

**Figure 4.8.2** – Area defined by location of the distal end of MC3 used for the data analysis. A) Areas defined as the parasagittal groove (PSG) and medial (M) and lateral (L) condyles associated with contact made by first phalanx (P1) - image credit Easton 2012. Areas included all data dorsal to the transverse ridge. B) Areas defined as the parasagittal groove (PSG) and medial and lateral condyles associated with contact made by the proximal sesamoids (PS). Areas included all data palmar to the transverse ridge.

Figure 4.8.3 – Mapped material stiffness on third metacarpal from micro-computed tomography images. Bottom images show the internal stiffness distribution through the frontal plane of the distal end of MC3.
Figure 4.8.4 – Material mapping and associated von Mises (MPa) stress distribution according to mesh density (coarse, moderate and fine) under impact loading in the same bone with varying mesh densities.
Figure 4.8.5 – Representation of different density-modulus equations used for human application and the equation used for this study (experimental) that was derived using specific data from material testing within the literature on the equine MC3. Maximum apparent density for the specimens scanned were 2.44 g/cm$^3$ and 2.47 g/cm$^3$ for the healthy and OA specimen respectively.

Figure 4.8.6a – Comparison of von Mises stress (MPa) on the distal aspect of MC3 in this study Healthy and OA (Top) to Easton 2012 static loading results for CTL - healthy and NFX – Osteoarthritic (Bottom)
M: Medial, L: Lateral, D: Dorsal, P: Palmar
Figure 4.8.6b – Comparison of average von Mises (MPa) stress between experimental static models and Easton (2012) FE models specified by location at 30° palmar on the distal end of MC3.
M: Medial, L: Lateral, CON: Condyle, PSG: Parasagittal Groove

Figure 4.8.7a – Distal end of the third metacarpal on the healthy finite element model, osteoarthritic finite element model and experimental pressure film indicating contact area and pressure (MPa) under impact loading at 3.55m/s. The greater the intensity of the red staining on the experimental pressure film indicates a higher contact pressure at that location.
Figure 4.8.7b – A comparison of the average contact pressures (MPa) across locations in the distal condyle between impact loading in the OA finite element model, healthy finite element model and the experimental ex vivo testing.
M: Medial, L: Lateral, PSG: Parasagittal Groove

Figure 4.8.8a – Comparison of von Mises (MPa) surface stress (left) and a lateromedial slice of von Mises stress (right) between healthy and OA models under impact loading.
M: Medial, L: Lateral
**Figure 4.8.8b** – Comparison of the average von Mises stress (MPa) at multiple locations on a lateromedial slice in the dorsal region of MC3 between the healthy and OA impact models. M: Medial, L: Lateral, SR: Sagittal Ridge, PSG: Parasagittal Groove

**Figure 4.8.9** – Comparison of the average contact pressure (MPa) by location on the distal end of MC3 between impact loading (OA, Healthy and Experimental) and static loading (Easton model trot, Easton model gallop and Easton experimental). M: Medial, L: Lateral, PSG: Parasagittal Groove
Figure 4.8.10a – Comparison von Mises (MPa) stress distribution in the distal aspect of the third metacarpal in the healthy and OA models under static and impact loading. Scales within loading conditions have been standardized for model comparison.

Figure 4.8.10b – Comparison of average von Mises stress (MPa) on a lateromedial slice in the palmar region of the third metacarpal across location between impact and static loading in the healthy and OA models.

Figure 4.8.10c – Comparison of average von Mises stress (MPa) on a lateromedial slice in the dorsal region of the third metacarpal across location between impact and static loading in the healthy and OA models.

Tables

<table>
<thead>
<tr>
<th>Model</th>
<th>Bone</th>
<th>Elements (#)</th>
<th>Surface Elements (#)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy MC3</td>
<td>219 232</td>
<td>18 914</td>
<td></td>
</tr>
<tr>
<td>Healthy P1</td>
<td>95 654</td>
<td>11 218</td>
<td></td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>314 886</strong></td>
<td><strong>30 232</strong></td>
<td></td>
</tr>
<tr>
<td>Osteoarthritic MC3</td>
<td>241 321</td>
<td>13 200</td>
<td></td>
</tr>
<tr>
<td>Osteoarthritic P1</td>
<td>90 508</td>
<td>10 664</td>
<td></td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>331 829</strong></td>
<td><strong>23 864</strong></td>
<td></td>
</tr>
</tbody>
</table>

Table 4.1 – Moderate mesh density details used for finite element analysis in the healthy and osteoarthritic impact loading simulations. Only the third metacarpal bone was used for the static loading simulations.

<table>
<thead>
<tr>
<th>Loading Type</th>
<th>Type</th>
<th>Boundary Conditions/number of nodes</th>
<th>P1/number of elements</th>
<th>PS/number of elements</th>
<th>MC3/number of nodes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Static (Standard)</td>
<td>Pressure (node)</td>
<td>Cut end MC3</td>
<td>10 MPa</td>
<td>12 MPa</td>
<td>n/a</td>
</tr>
<tr>
<td>healthy</td>
<td>2718</td>
<td>2962</td>
<td>1393</td>
<td>n/a</td>
<td></td>
</tr>
<tr>
<td>OA</td>
<td>1981</td>
<td>2046</td>
<td>927</td>
<td>n/a</td>
<td></td>
</tr>
<tr>
<td>Impact (Explicit)</td>
<td>Velocity (node)</td>
<td>Cut end P1</td>
<td>n/a</td>
<td>n/a</td>
<td>3.55 m/s</td>
</tr>
<tr>
<td>healthy</td>
<td>2369</td>
<td>n/a</td>
<td>n/a</td>
<td>279 182</td>
<td></td>
</tr>
<tr>
<td>OA</td>
<td>1740</td>
<td>n/a</td>
<td>n/a</td>
<td>336 844</td>
<td></td>
</tr>
</tbody>
</table>

Table 4.2 – Number of nodes and elements used to specify boundary conditions and loading in each model.

<table>
<thead>
<tr>
<th>Resolution</th>
<th># elements MC3</th>
<th># elements P1</th>
<th>Total # elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coarse</td>
<td>20 611</td>
<td>13 294</td>
<td>33 905</td>
</tr>
<tr>
<td>Moderate</td>
<td>241 321</td>
<td>90 508</td>
<td>331 829</td>
</tr>
<tr>
<td>Fine</td>
<td>1 197 231</td>
<td>406 604</td>
<td>1 603 835</td>
</tr>
</tbody>
</table>

Table 4.3 – Number of elements in each instance for the 3 mesh densities used to determine convergence.

<table>
<thead>
<tr>
<th>Resolution</th>
<th>Medial Condyle</th>
<th>Medial PSG</th>
<th>SR</th>
<th>Lateral PSG</th>
<th>Lateral Condyle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coarse – Fine</td>
<td>0.207</td>
<td>0.37</td>
<td>0.129</td>
<td>-0.034</td>
<td>0.153</td>
</tr>
<tr>
<td>Fine</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Moderate – Fine</td>
<td>0.050</td>
<td>0.036</td>
<td>-0.024</td>
<td>-0.043</td>
<td>0.041</td>
</tr>
<tr>
<td>Fine</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 4.4 – Difference in average von Mises stress by location between the selected model (moderate) and the fine and course resolution models. PSG: Parasagittal Groove, SR: Sagittal Ridge.
4.9 Appendices

Appendix A

Appendix A - Cross-sectional slices of von Mises (MPa) stress distribution by location for both models under each loading condition.
P: Palmar, D: Dorsal
Appendix B – 160 point equidistant map used for von Mises (MPa) stress analysis. Figures below are the plotted stress (MPa) results from the 160 point map (rows 1-10 and columns C1-C16) based on slice (palmar/dorsal), model (healthy/static) and loading condition (static/impact).
OA Static - dorsal

Lateral

Medial

Stress

C1
C3
C5
C7
C9
C11
C13
C15
C17
C19
C21

proximal

distal

1

7
SUMMARY AND FUTURE DIRECTIONS

These studies provide unique and original findings on the effect of impact loading within the equine metacarpophalangeal (MCP) joint and provide the framework for future investigations into the characterization of impact loading and the potential role this modality of loading may have on injury within this joint. While impact loading has been previously evaluated (Gustås et al. 2006, Thomason et al. 2008), the effect of impact loading on joint contact within the MCP and resulting stress distribution within third metacarpal (MC3) remained unknown. The relationship of impact loading (high magnitude, high frequency) and the deleterious effect it has on bone in association with joint disease has been recognized for over 40 years (Radin 1972). Even so, the high forces associated with midstance loading have been the primary focus within equine research in determining the relationship between biomechanical loading and osteoarthritis (OA) within the MCP joint of performance horses. This current work suggests that impact loading, in addition to midstance loading, should be considered for future study in the context of injury based on the high shock loading that results in subsequent bone stresses that are comparable to those found in midstance.

The first study examined the effect of hoof angle at impact and the resulting accelerations. Although the hoof acts as an effective shock absorber, there appears to be significant shocks reaching up into MC3 suggesting that the structures from hoof to mid-diaphysis MC3 are subject to high frequency, high magnitude loading with every foot fall of the horse. There is a lack of evidence within the literature to determine the effect of high frequency signals on bone tissue and further work is needed in this area.

The second study provided insight into the contact area within the MCP joint under impact loading. The proximal sesamoids (PS) did not play a significant role during impact with
the primary contact area being associated with the loading from the first phalanx (P1). The greatest contact area and concentration of high pressures were found on the dorsal aspect of the medial condyle, an area found to be associated with the greatest bone density in the distal end of MC3 (Young et al. 2007). It is possible that this high density region is the response to receiving high magnitude loading under both impact and midstance phases. To the best of the authors’ knowledge, this was the first study to examine the joint contact area at primary impact, however further work is needed in this area in order to determine the maximum joint contact pressure within the MCP joint under a wider range of impact velocities.

The final study created two finite element models (FEMs) that was validated using the joint contact areas and pressures from the existing literature and from data in the second study. While other FEMs of the equine distal limb structures have been created (Les et al. 1997, Hinterhofer 1997/2001, McClinchey 2003, Collins 2009, O’Hara 2012, Easton 2012), none have loaded these models under impact conditions. The use of FE modeling provides an opportunity to compare different loading modalities and magnitudes within the live horse during multiple phases of the stance. The results of impact loading compared to static (midstance) loading within the final study showed the dorsal aspect of MC3 (where loading occurs both in impact and midstance) to have similar stress magnitudes, even slightly greater under impact loading. Although the mass involved under impact loading is much less (5kg verses 2.5 times the horses bodyweight at midstance), the high accelerations occurring during this phase have shown to produce significant bone stresses within MC3.

Understanding the role of biomechanical loading and the potential effect it had on the musculoskeletal system provides useful information to the equine industry. Evidence-based knowledge can be used to determine areas where improvements can be made in order to
minimize injuries (i.e. footing, training programs). Footings research is currently ongoing, however this area is challenging as the properties that are ideal for one phase of the stance, may be worse for another. Determining the ideal conditions through characterization of each phase of the stance can provide useful information to determine the optimal footing for performance horses. Based on the results from this body of work, it is evident that impact is an important factor to consider in reducing the effect of biomechanical loading on the distal forelimb of performance horses. Continuing work within this area will hopefully provide the evidence needed to establish the link between biomechanical loading and the etiology of OA, allowing the equine industry to focus on areas to improve animal welfare by decreasing the incidence of bone and joint disease.
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