# QUANTIFYING THE EFFECT OF SAGITTAL PLANE LOADING ON ANTERIOR-POSTERIOR AND COMPRESSIVE KNEE LOADS DURING DYNAMIC SIMULATIONS

By

Fallon G. Fitzwater

Submitted to the graduate degree program in Mechanical Engineering and the Graduate Faculty of the University of Kansas in partial fulfillment of the requirements for the degree of Master of Science.

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Chairperson Lorin Maletsky

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Carl Luchies

_____

Ken Fischer

Date Defended: Oct 30, 2014

The Thesis Committee for Fallon G. Fitzwater

certifies that this is the approved version of the following thesis:

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Chairperson Lorin Maletsky

Date approved: Dec 05, 2014

## Abstract

Dynamic in-vitro knee simulator's ability to achieve physiological compression and A-P loads is dependent on the chosen axes of applied loads in the sagittal plane. Anterior-posterior (A-P) and compressive forces generated at the tibia during dynamic activities have a large influence on the joint kinematics and soft tissue loads. Simulating accurate physiological joint forces during in-vitro testing is essential for evaluating the performance of total knee replacements and determining the effects of pathologies and injuries. Instrumented tibia trays have measured joint forces for a variety of activities. Most activities generate compressive loads that range from 2-3.5 times body weight (BW). A-P forces range between a .40 BW posterior to a .20 BW anterior. The objective of this study is to determine the relationship between hamstring force and ankle load, which cause a flexion-extension moment at the ankle, and A-P and compressive tibia forces in a dynamic knee simulator.

A set of posterior stabilized total knee prostheses attached to custom fixtures was mounted on the Kansas Knee Simulator. A non-physiological deep knee bend, ranging from 20-120° flexion, was performed while applying constant ankle load and hamstring forces independently. Joint forces were measured using a six degree-of-freedom load cell placed directly below the tibia tray. A-P translations of the femur relative to the tibia were calculated for each cycle. In addition, a rigid body computational model with either a hamstring force or ankle loads (ADAMS.MSC software 2011) was used to determine the five loading profiles for the Kansas Knee Simulator needed to simulate a physiological walk compression and A-P loads obtained from instrumented tibia data.

The addition of hamstring forces increased the anterior forces at the tibia while increasing compression. A posteriorly directed ankle load, which causes an extension moment at the ankle, caused a decrease in anterior forces but did not have the capability to generate a posterior

directed tibia force. Posterior ankle loads decreased quadriceps loads and compression. In the computational model, the hamstrings were not able to achieve posterior tibia loads while an anterior ankle load is required to achieve posterior joint force during the walk simulations.

Dynamic knee simulators in which the hip positioned directly over the ankle during the entire simulation can easily achieve anterior tibia loads but will have difficulties generating a posteriorly directed tibia load. The hamstrings have shown to only contribute to anterior tibia loads and will not be necessary to include in this type of knee simulator. Anterior ankle loads can create posterior tibia loads; however, it is highly coupled with the quadriceps axis which is typically run in position control. The anterior ankle loads, used to create a posterior tibia load will increase quadriceps loads and compression. The results from this study show that ankle loads, which cause flexion-extension moments at the ankle, are a robust way to achieve a physiological ratio of A-P to compressive load at the tibia.

## Acknowledgements

I owe special thanks to my advisor, Dr. Lorin Maletsky, for the incredible opportunity to become a member of EJBRL. I appreciate all of your honest and critical feedback which has helped me grow as both as a person and as an engineer.

I appreciate my other committee members, Dr. Carl Luchies and Dr. Ken Fischer, for their participation in this work. I would also like to thank all other professors in the School of Engineering.

I would like to thank EJBRL Lab members past and present, Sami Shalhoub, Adam Cyr, Max Eboch, Lauren Ferris, Kaity Morrow, and Amber Lenz who contributed to this work. This thesis would not have been possible without your invaluable feedback and assistance. I would like to especially thank Amber Lenz and Mike Strickland who developed the original model used in this study.

I would like to thank all of family members Bridget, Montgomery, Sara, Ian, Leah, Ry, Rem, Will, Ella and Maddie Fitzwater for their love, support, and for their understanding of this time commitment.

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## List of Abbreviations

| | | |
|---|---|---|
| ACL | | Anterior Cruciate Ligament |
| AL | | Ankle Loads |
| | aAL | Anterior Ankle Loads |
| | pAL | Posterior Ankle Loads |
| A-P | | Anterior-Posterior |
| BW | | Body Weight |
| DKS | | Dynamic Knee Simulator |
| I-E | | Internal-External |
| KKS | | Kansas Knee Simulator |
| LP | | Lowest Point |
| M-L | | Medial-Lateral |
| TKA | | Total Knee Arthroplasty |
| TF | | Tibiofemoral |
| VF | | Vertical Force |

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## 1. Introduction

Instrumented tibiae have allowed for accurate and reliable measurement of anterior-posterior (A-P) and compressive tibia loads generated at the knee for multiple activities of daily living [1-6]. While variability exists in measured tibia loads across subjects for a given activity the ratio of A-P to compression is highly dependent on the activity performed [1]. Most activities create both anterior and posteriorly directed tibia loads at some portion of the activity cycle. A-P forces range between a posterior .40 BW to an anterior .20 BW, which is 10-20 times smaller in magnitude than compression. The ratio of A-P to compression generally lies in a narrow range of .5 and -.5 for most activities of daily living [1].

Applied external loads in the sagittal plane will likely have the most influence on A-P and compressive joint forces. The sagittal plane loads also have a significant effect on the tibiofemoral flexion moment. The complex and coupled interaction between multiple externally applied loads can make it challenging to determine the individual contribution of an applied load to A-P and compressive knee forces.

Dynamic in-vitro knee simulators (DKS) are used extensively for biomechanical evaluation of the human knee. Physiological joint forces generated during activities are the net result of bodyweight loads, ground reaction forces and muscle loads. DKS aim to replicate in-vivo joint forces and tibiofemoral fleixon though a simplified loading configuration. External loads are typically applied to bone ends and select musculature while allowing unconstrained kinematics at the joint [7, 8].

The chosen axes of applied loads and selected loaded musculature will be a deciding factor in a DKS rig's ability to simulate a given activity. Choosing axes of applied external loads and

creating dynamic loading profiles for DKS can be challenging, therefore, it is advantageous to use a single actuator to influence a single physiological direction load while having minimal effect on the other physiological loading directions. This un-coupling of actuators will likely make control and development of loading profiles easier for dynamic simulations. Evaluation of the individual contribution of applied external loads to resultant joint loads will aid in determining which axes of applied loads are essential to creating a physiological A-P and compressive joint forces.

The current study has two main objectives. The first objective it to quantify the effects of externally applied sagittal-plane loads, hamstrings and ankle loads, on compression and A-P knee kinetics and kinematics and quadriceps loads. The second objective is to determine the magnitude and direction of loading profiles necessary to simulate a physiological walking profile, containing both anterior and posterior knee loads. This study will identify if an external load contributes to either anterior or posterior tibia loads. Determining applied loads that are necessary to generate a physiological A-P and compressive joint load can improve dynamic simulations.

The literature review in chapter 2 will describe loading conditions of the knee and how these loads are measured during functional activities. It will also describe previous studies, which aimed to determine the effects of sagittal plane loads A-P kinematics, kinetics and soft tissue structures of the knee. Finally, it will describe the dynamics knee simulators and computational model used in this study. Chapter 3 will explain the effects of applied loads on resultant kinetics and kinematics and present loading profiles necessary to simulate a physiological walking cycle. Chapter 4 will provide a conclusion of this work and present recommendations for how this study can improve the in-vitro simulations using dynamics knee simulators.

## 2. Literature Review

This literature review will describe joint forces generated at the knee during various activities of daily living and how these forces are measured in in-vitro studies. This chapter will also review previous studies, which aimed to determine the contribution of external loads in the sagittal plane to anterior-posterior (A-P) kinetics, kinematics and loading of soft tissue structures.

### 2.1 Determining In-Vivo Knee Loads

Previously, in-vivo dynamic knee loads were predicted using gait measurements, predicted muscle forces, and articular geometry [9, 10]. These studies typically used a scaled musculoskeletal model and inverse dynamics to predict joint forces. The resultant joint forces from these musculoskeletal models were highly dependent on model parameters and assumptions. Uncertainties in parameters in the inverse dynamic simulations likely caused variation in predicted knee loads found in the literature. Wehner et al. estimated a maximum axial load of 470% body weight (BW) during normal walking [11]. Other studies reported maximum axial loads during normal gait of 600% BW and 300% BW by Glitsch et al. and Taylor et al. respectively [10, 12]. Kuster et al. reported the highest axial load of 800% BW during downhill walking [13].

To overcome the inconsistencies in predicted knee loads from literature and to obtain direct in-vivo tibiofemoral loads during activities, compartmental instrumented tibiae and instrumented total knee replacements were developed. Taylor et al. was the first to attempt to calculate in-vitro tibiofemoral joint loadings using an instrumented proximal femoral implant, with strain gauges mounted along the femoral stem approximately 200 mm above the joint line [6]. Taylor et al. measured a maximum axial load of 230% BW during walking [6]. More recently, D'Lima et al. developed a more advanced instrumented total knee replacement that directly measured in-vivo

tibia loads [4, 14]. This instrumented prosthetic design also utilized strain gauges placed along the distal tibia stem. A telemetry system amplified and transmitted data to an external receiver [4]. The measured peak tibia loads during normal gait were in the range of 280% BW. Kutzner expanded on available instrumented load cell data by collecting a variety of daily living activities for multiple subjects [5]. This study obtained similar peak axial knee loads as D'lima et al. and Taylor et al., with 261% BW peak axial load for walking. These measured loads using instrumented components reported much lower loads than was previously calculated loads using inverse dynamics methods. In addition, the reported measured loads are more consistent for a specific activity between different studies and instrumented prosthetic designs. These studies had subjects perform various daily living activities, including stair ascent, stair descent, sitting down, standing up, deep knee bends, and even light athletic activities such as golf swing. Measured tibiofemoral loads showed variations within a single subject (intra-individual) and between subjects performing the same activity (inter-individual) [5]. Kutzner reported an overall intra-individual variation range of 50% BW for level walking for a single subject. The inter-individual variation range was slightly higher at 70% BW across all five subjects for level walking. The activity performed also affected inter- and intra- variations. The knee bend had the highest inter-individual variation with a range of 118% BW.

Instrumented tibia data suggests that the activity performed greatly affects the percent BW loads generated at the knee. Investigating loads at the knee as a percentage of BW is a way to normalize across subjects and minimize inter-subject variability. Although there are consistent loads as a percent of BW generated at the knee for a specific activity, there is still a range of loads that span the variation between subjects. Simulation of a specific activities in a dynamic

knee simulator can either reflect the in-vivo loads of a specific subject or include a range of simulations to capture the variation between subjects.

## 2.2 Anterior-Posterior Kinematics

In-vivo knee kinematics have been widely documented for a variety of activities [15, 16]. Reported A-P kinematics have some variability between studies due to the methodology used for data collection, the type of articular geometry, and the technique used for calculating A-P motion. Fluoroscopy is a common method used to determine in-vitro kinematics. Video fluoroscopic images frames are registered to precisely match computer generated three-dimensional models of the subject's knee geometry created from MRI images using an optimization algorithm. Anterior-posterior knee joint motion has been previously described using Grood-Suntay defined coordinate system for describing joint motion [17]. The A-P axis is described in this coordinate system as floating axis that is not fixed to either the femur or tibia. A-P kinematics described using Grood-Suntay does not provide direct A-P translations. Lowest point (LP) is an alternate method for describing A-P motion that tracks translation of the lowest point of the femur along the direction of the long axis of the tibia. Tracking of the contact point of the femur on the tibial insert gives a better description of A-P motion of the knee. In addition, the LP can be separated into translations on both the medial and lateral tibial plateau. Average A-P motion can potentially be masked by internal and external (I-E) rotations of the tibia using the standard kinematic description, but by describing the medial and lateral translations separately, condylar A-P motions can be examined along with IE rotations. Studies have documented kinematics for both subjects with natural knee geometry and with multiple designs of total knee arthroplasty, (TKA) [15, 16, 18]. These studies determined that the articular geometry and the activity performed have a large influence on A-P translations.

## 2.3 Anterior-Posterior and Compressive Knee Loads and Kinematics during In-Vitro Simulations

The A-P and compressive knee loads are important due to their direct effect joint kinematics and ligament loads. Anterior cruciate ligament (ACL) injuries are a common debilitating knee injury with approximately 200,000 cases per year in the United States alone [19]. Multiple studies have established that strain in the ACL is directly related to A-P loads at the knee during weight bearing activities [20-23]. Fleming reported a large increase in ACL strain for an applied A-P load when transitioning from non-weight bearing to weight bearing activities. Anterior translations are linked to both anterior knee loads and to increase weight-bearing acceptance [24]. This suggests that the ratio of A-P to compression has a significant effect on soft tissue loadings and A-P kinematics. Accurate simulation of the physiological ratio of compressive load to A-P load is critical for achieving physiological A-P kinematics, soft tissue loads and timing of post cam engagement during in-vitro simulations. Sagittal plane loading conditions of DKS can include a vertical force, muscle forces and moments about the ankle that are directly related to the ratio of SI to A-P loads generated at the knee during simulations. D'Lima et al. was able to accurately predict I-E and A-P kinematics with an accuracy of .5° and .5 mm [2].

In-vitro studies have also studied how external loading conditions affect A-P kinematics. In-vitro and computational studies have shown that quadriceps dominated simulations, simulations that solely use a quadriceps mechanism to control tibiofemoral (TF) flexion, resulted in kinematics that had high anterior femur translations, high anterior tibia knee loads, and high anterior cruciate ligament (ACL) strain [21-23, 25, 26]. In addition, the use of an antagonistic hamstring load resulted in a decrease in anterior tibia forces or even created posterior tibia loads, and greatly decrease ACL ligament strain [22, 23, 25, 26]. Understanding the effects of quadriceps and

hamstring loads on joint forces can provide valuable information on how to create physiological A-P and compressive joint loads during in-vitro simulations. The coupled interaction between applied external loads is also important to consider during dynamic simulations.

## 2.4 In-Vitro Dynamic Knee Simulations

In-vitro knee testing allows for comparative evaluation between healthy natural knees and subsequent changes to the joint such as simulated injury, joint replacement, or surgical alignment for a single specimen [18, 27]. Knee testing rigs are used extensively for the biomechanical evaluation of the knee. Numerous knee-testing rigs have been constructed to apply loads and motions to the knee joint through either bone ends or selected musculature. The capabilities of these machines depend on chosen degrees of freedom, applied loads, and simulated musculature.

Quasi-static rigs typically position the knee at a desired flexion angle while applying external loads. The robust configuration of quasi-static rigs allows for complex loading conditions and the ability to load multiple muscle that span the knee. Quasi-static rigs are limited in their ability to dynamically simulate physiological activities. Dynamic knee simulators can be classified into two basic categories wear simulators, and physiological simulators. Wear simulators are used to assess the generation of wear particles created through cyclic loadings of prosthetic components [28-30]. Flexion angles and cyclic duration are performed during wear simulations according to ISO 14243 under either loading (ISO 14243-1) or displacement profiles (ISO 14243-3). Wear simulators mount prosthetic components into seal temperature controlled chambers which replicate the environment of the human body. Wear simulators are not used evaluate knee pathologies or surgical techniques as they do not include soft tissue structures. Typically, these wear simulators only have the ability to determine prosthetic components wear and failure characteristics.

Dynamic knee simulators (DKS) attempt to replicate a physiological loading condition at the knee for a given activity. DKS apply external loads and a flexion profile while allowing unconstrained fine kinematics at the joint [8]. The well-controlled and repeatable loading environment of these simulators allows kinematic changes to be attributed to changes in soft tissue structures and geometry alone. These types of rigs are especially useful for evaluating prosthetic designs as they allow for direct kinematic comparison to natural geometry for a single specimen [18, 27]. Previously, input loads used to drive DKS were obtained from direct measurement of gait study parameters such as ground reaction forces, kinematics, and muscle activations. Currently, instrumented tibiae provides detailed and accurate knowledge of in-vivo joint forces which is used along with computational models to calculate input loading profiles required to simulate a physiological activity.

DKS have a variety of design and loading configurations, methods of controlling TF flexion, and joint load capabilities. The Oxford rig has been used extensively for biomechanical testing of knee and additional rig have been creating using the Oxford rigs basic configuration [7, 8, 21, 31]. This literature review focus on describing knee simulators based on this configuration. In these types of rigs, the knee is upright with the hip positioned directly over the ankle. Flexion-extension (FE) and vertical translations are allowed at the hip while the ankle assembly allows FE and Ad-Ab and I-E [8]. A position controlled quadriceps mechanisms controls knee flexion angle by balancing the flexion moment at the knee.

Variations to this basic rig design can include additional musculature and applied external loads. Muller et al developed the Tuebingen knee simulator based on the Oxford rig basic design but separated the quadriceps into three groups and included two heads of the hamstring [31]. The muscle bodies were clamped and attached to electrical servo motors. In this rig, TF flexion was

controlled by hip translations and muscle forces were applied to obtain a compressive joint load, which is measured by load cells at the hip and ankle joint bearings. The Tuebingen rig design does not attempt to control A-P loads at the knee. The Munich dynamic knee rig also mimics the Oxford rig configuration [32]. The vertical force was used to create a compressive load across the knee. The quadriceps loads are ran at a constant displacement rate to control the flexion angle. Hamstrings were loaded with a constant load distributed equally between the medial and lateral hamstring loadings. A tri-axial load cell mounted directly below the tibia measures knee loads. This rig did not attempt to generate a specific A-P joint condition.

The Kansas Knee Simulator (KKS) is a 5-axis hydraulically controlled knee simulator that aims to replicate dynamic physiological loading conditions at the knee [7]. The KKS is designed to apply external loading conditions to either a cadaveric knee specimen or prosthetics mounted onto custom fixtures, without constraining knee kinematics, allowing for movement in all 6-degree of freedom at the joint. Three axes that lie in the sagittal plane include, a vertical axis at the hip, ankle force axis applied at the ankle, and a quadriceps mechanism. Two additional out of sagittal plane axes are the I-E and Ad-Ab axes at the ankle. All five axes have the capability to be ran in either load or position controls. The KKS controller utilizes a five-channel PID controller with full cross-compensation and receives feedback from uniaxial load cells and linear and rotary displacement transducers mounted on the rig.

Loading profiles created for the five axes of the KKS aim to replicate in-vivo loading conditions and knee flexion obtained from instrumented tibia data. The loading profiles are calculated using a computational model of the KKS which is described in detail in the next chapter.

## 2.5 Computational Model of the KKS

The Kansas Knee Simulator has been modeled in ADAMS.msc software which accurately reflects rig geometries and inertial properties. The knee is modeled using CAD geometry from a TKA. The TF contact force is defined using Hertzian contact between two elastic bodies, which calculates a normal force based on the penetration depth determined by the separation between two bodies surfaces, the material modulus of elasticity, and Poisson's ratio. To simplify the geometry in ADAMS, each condyle was approximated using elliptical spheres. The quadriceps mechanism accurately reflects the Kevlar strap used in the physical rig and includes wrapping effects by including several "points" that generate a contact force to the sagittal plane geometry of the femoral component that is modeled with a simple "polyspline". The model has been instrumented with a triaxial load cell marker placed in the same location as the load cell on the physical rig.

The purpose of this model is to generate loading profiles of the KKS for various activities. The model uses loads at the knee and gross TF flexion as inputs to the KKS plant and then calculates the loading profiles for the five actuators (Fig. 1). This process is facilitated using an advanced PID control system implemented in Simulink. The Simulink controller schematic for the model with the hamstring inclusion can be found in Appendix A. The ADAMS model of the KKS is exported as a system of state space equations and incorporated into the Simulink system as the "plant" subsystem. Inputs to the KKS subsystem are loads of all five actuators. The outputs to the plant are feedback from loads at the knee, and position and loads of all the five axes of the KKS. Load limiting and slew-rate factors are included on each actuator to limit loads and accurately reflect delays in the hydraulic system.

All actuators included frictional and dampening effects and measure loads and displacements consistent with sensors on the rig. The hamstring actuator was included in this model. The actuator housing was mounted to the hip sled assembly. The actuator rod was attached to a single attachment site on the posterior side of the tibia to maintain a line of action parallel to the long axis of the femur. This new actuator was included into the plant subsystem as an additional input. This allows for predicting hamstring loads necessary to achieve a specific joint loading condition.

This additional hamstring axis was evaluated by running both the model and the simulator with a constant hamstring load of 100 N on a walk cycle (Fig. 2). The model was in good agreement with quadriceps and compressive joint forces. The anterior loads had similar magnitude but differed in their profiles. This difference in anterior load is likely a result of slight variations in applied hamstring loads in the experimental setup.

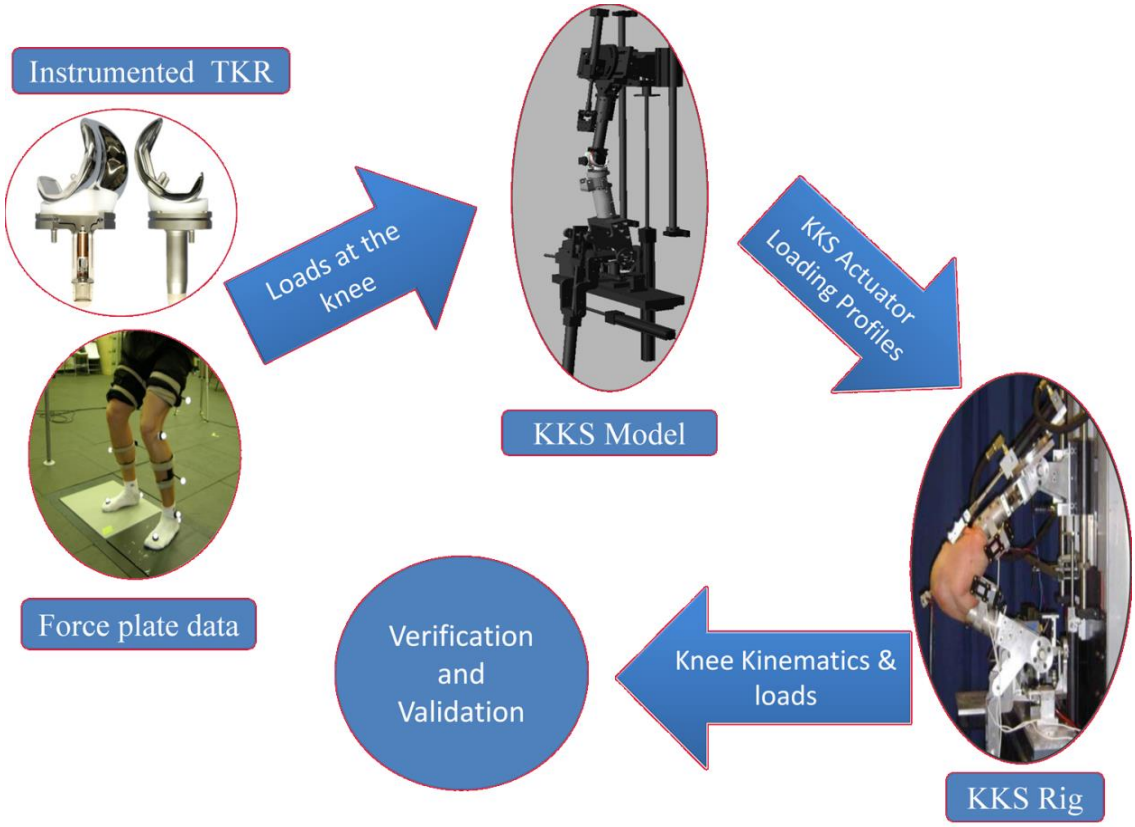


**Figure 1: Flow chart of the profile generation process used to generated loading profiles for dynamic knee simulators using a computational model**

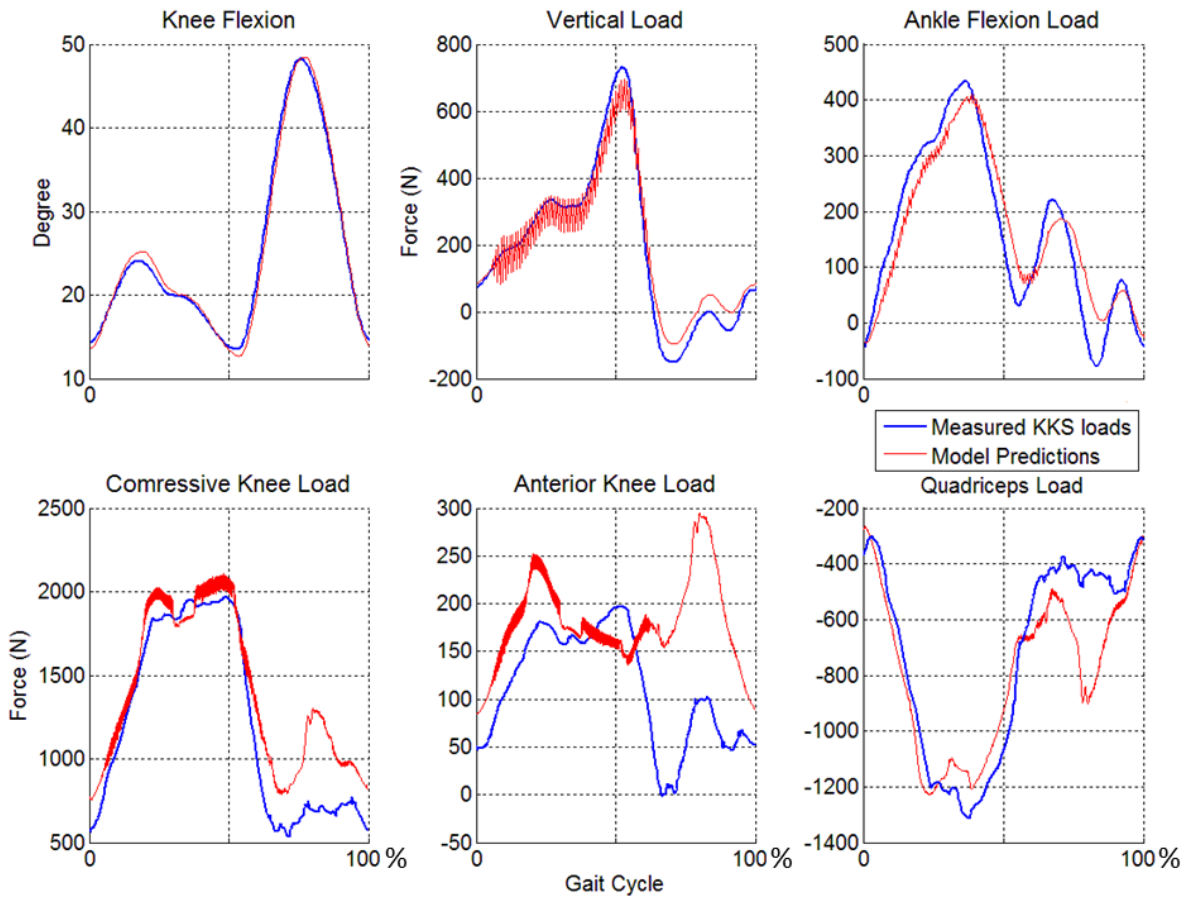


**Figure 2: Comparison between KKS model predictions and the KKS tracking both with the inclusion of a constant hamstring load of 100 N**

# 3. Journal Article: The Influence of External Loading Conditions on Anterior-Posterior and Compression Loads of the Knee

## 3.1 Introduction

Compressive and anterior-posterior (A-P) loads at the tibia have been measured in-vivo using instrumented tibiae for multiple activities of daily living [1-3, 14, 33]. For most of these activities, the compressive load at the knee ranged from 2 to 3.5 times bodyweight (BW) while A-P loads were approximately .1 to .3 times body weight generally 10 to 20 times smaller in magnitude than the compressive force. The ratio of compression to anterior-posterior (A-P) knee joint loads generated during physiological activities have substantial effect on the loading of soft tissue structures, patellofemoral loads, and knee kinematics [20-26, 34]. D'Lima et al. observed a direct relationship between A-P and compressive joint forces and resultant A-P kinematics [2]. The researchers confirmed that the direction of A-P load at the knee was associated with the direction of knee kinematics: an anterior load measured on the tibia resulted in an anterior translation of the femur relative to the tibia. D'Lima et al. was also able to compute A-P kinematics with an error of less than 1.2 mm from measured instrumented tibiae A-P and compression tibia loads, at a given knee flexion angle.

The net compressive and A-P loads at the knee are the result of multiple applied external loads such as muscle forces, ground reaction forces, and bodyweight loads. Multiple studies have investigated the effect of changes in these external loads on A-P and compressive joint forces. Musculoskeletal models using a EMG-driven approach determined that hamstring forces reduce anterior tibia forces while quadriceps and patella tendon loads contributed to anterior tibia forces

[22, 26]. Dynamic in-vitro knee simulations have also determined that hamstring forces cause decreases in anterior tibia loads and can even create posteriorly directed loads [21]. Additional cadaveric studies have determined that increases in quadriceps forces cause increased anterior tibia forces [21, 23]. Other studies determined that quadriceps and hamstrings were not significant indicators of knee joint forces [25, 26]. The conflicting results found in the literature are likely a result of varied measurement techniques or the locations where tibia loads are measured. To make meaningful comparisons to instrumented tibiae, which only measure the femur contact forces, in-vitro testing should measure knee forces in the same manner. Predicting the magnitude and direction of applied loads necessary to create a physiological joint loading condition will allow in-vitro studies to accurately simulate activities of daily living. The first objective of this study is to quantify the effects of posterior ankle loads (pAL), which cause and extension moment at the ankle and hamstring forces on A-P and compressive tibia loads. The second objective is to calculate the applied loads on an in-vitro knee rig that are necessary to replicate the A-P and compressive tibia loads of a physiological gait cycle, using either a hamstring load or an ankle load.

## 3.2 Material and Methods

A posterior stabilizing total knee replacement (TKR), P.F.C Sigma (DePuy Synthes, Warsaw, IN), was mounted on the Kansas Knee Simulator (KKS) using aluminum fixtures (Fig. 1). The KKS is a 5-axis servo-hydraulic dynamic knee simulator that has the ability to simulate physiological loading conditions on a knee (Figure 3) [7]. The five axes of the simulator include three in the sagittal plane: a vertical force (VF) applied as a hip load, ankle load (AL) that creates a flexion-extension moment about the ankle, and a quadriceps load used to control knee flexion angle. Medial-lateral (M-L) loads and internal-external (I-E) torques are applied to the tibia at

the ankle. The KKS has a five channel PID controller which receives feedback from single axial load cells and position sensors from each of the five axes. All five axes have the ability to be run in either position or load control. The quadriceps axis is typically run in position control to track a knee flexion angle profile. The quadriceps actuator applies a load along a physiological straight line of action parallel to the long axis of the femur approximating the position of the rectus femoris. The M-L axis creates a vargus-valgus moment at the knee and the I-E axes creates a torque about a vertical axis. Inputs to the M-L and I-E axis are representative of ground reaction forces that can be obtained directly from force plate data. The ankle load axis has the ability to create either a flexion or extension moment at the ankle. A posterior ankle force (pAF) creates an extension moment while an anterior ankle force (aAF) causes an flexion moment at the ankle. Coupling exists between the five axes especially in the sagittal plane. In this rig configuration a single axis does not solely contribute to an isolated physiologically directed tibia load. The resultant joint load is the net contribution of all five axes. The goal of the loading profiles for all five axes of the KKS is to achieve targeted physiological joint forces.

A simple non-physiological profile was used to quantify the effect of externally applied loads on knee kinetics. This profile consisted of a deep knee bend, which flexed the knee from 20-120° and did not include any out-of-sagittal plane loading. The vertical loading profile in combination with the quadriceps load maintained a compressive load on the knee throughout the entire cycle. During early flexion the VF was applied downward force to maintain compression while during deep flexion the VF was applied an upwards force to help limit excessive quadriceps force.

This squat profile was performed for multiple iterations with different constant pAL, spanning 0 to 100 N in 10 N increments. In addition, the squat was simulated for multiple constant applied hamstring loads (97N, 147N, 220N, 257N). The hamstring load was applied using constant force

springs attached at the hip assembly and applied a posterior load to a single attachment point on the posterior side of the tibia. The hamstrings maintained a line of action roughly parallel to the long axis of femur. The values for hamstring forces were chosen based on the availability of constant force springs.

A tri-axial load cell (JR3, CA) mounted below the tibial tray recorded the loads applied on the tibia during simulations. Feedback tracking from each actuator was recorded during each cycle using Instron Wave Matrix software. Kinematics were recorded using an Optotrack Certus motion capture system (Northern Digital, Ontario). Anatomical landmarks on the tibia and femur prosthetic components were digitized to describe the kinematics using a three-axis orthogonal coordinate system [17]. A hexahedral mesh was generated for the solid model of the femur and tibia geometries using Hypermesh software (Altair, Alabama). The femur geometry was then transformed into the tibial coordinate system at every point of the cycle. The lowest point (LP) was calculated by determining the most inferior point of the medial and lateral femoral condyles along the superior-inferior (SI) tibial axis.

A computational model of the KKS rig was created in the multibody dynamic software Adams (MSC.ADAMS 2011) to accurately reflect the mechanical properties of the rig. Each of the five axes were represented using actuators that included position and load sensors in the same locations as the physical rig. The model was included in a basic feedback PID controller co-simulated in MATLAB/Simulink which uses loads at the knee and flexion angle as inputs, and calculates the actuators' loading profiles necessary to achieve the joint loading conditions at the knee. The joint loads as a percent of bodyweight from eight subjects performing a walking cycle obtained from the Orthoload database, were averaged across subjects [1]. Compression and A-P joint loads as well as TF flexion were used as inputs to the model (Fig. 2) [1]. Compression and

A-P loads as a percentage of body weight measured during the walking cycle obtained from the instrumented tibia showed moderate variability, 80% BW between the six subjects with posterior tibia loads generated during the early part of the gait cycle and anterior loads during the swing phase. The hamstrings load was included in the model with the actuator attached at the same location as the experimental setup. The Adams model calculated the actuator input profile loads for two configurations of the model. The first configuration included vertical force, quadriceps, and ankle force axes. For the second configuration maintained the vertical force and quadriceps and included an additional hamstring axes without the ankle force axis.

The model results from the hamstring and ankle configuration was investigated to determine the advantages and disadvantages of each configuration and their ability to achieve the targeted physiological input joint loads. The magnitude and direction of the actuator loads along with their influence on other actuator loads such as the quadriceps was compared for the two configurations.

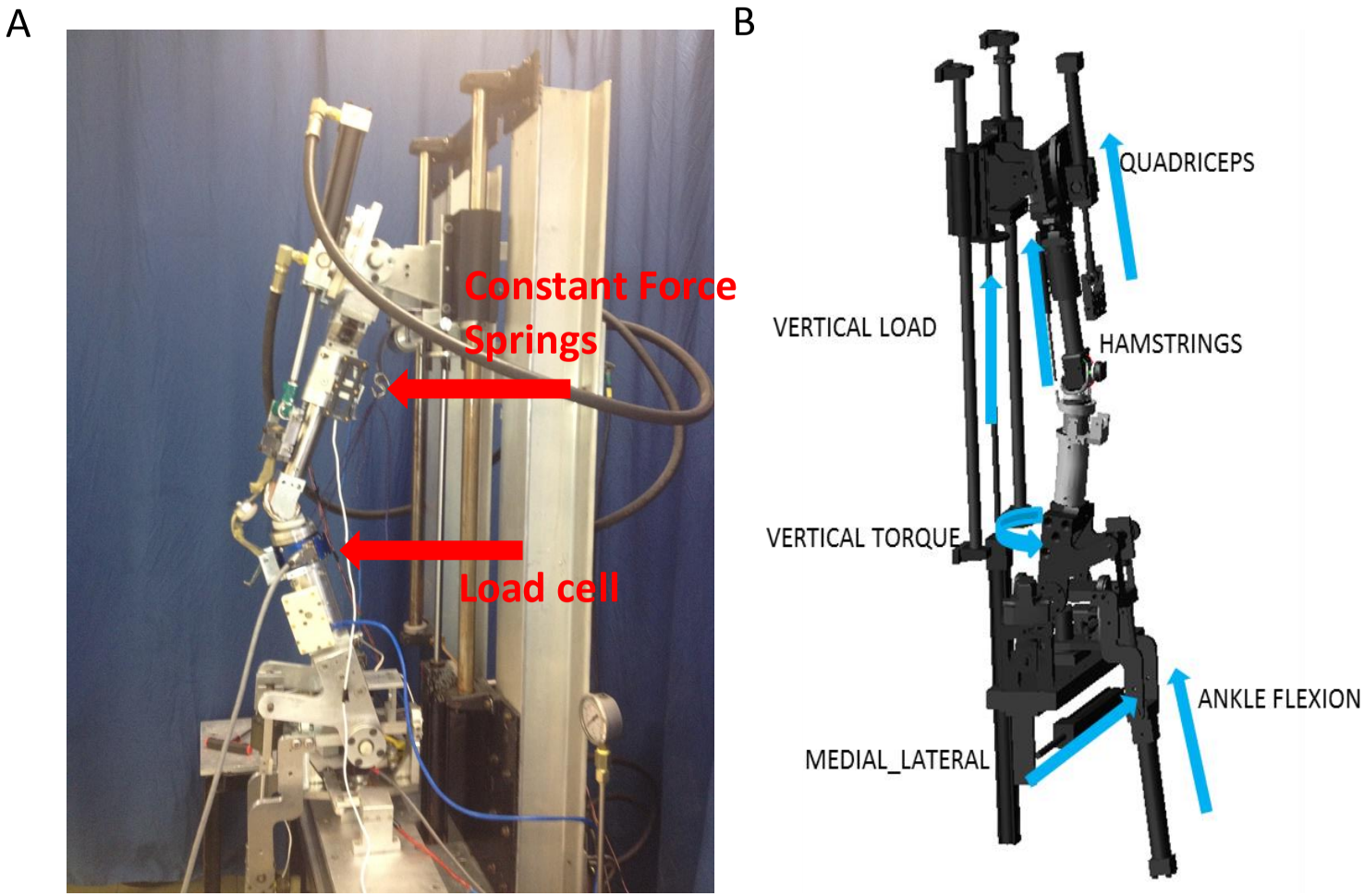


**Figure 3: The experimental KKS rig (A) and model of the KKS (B)**

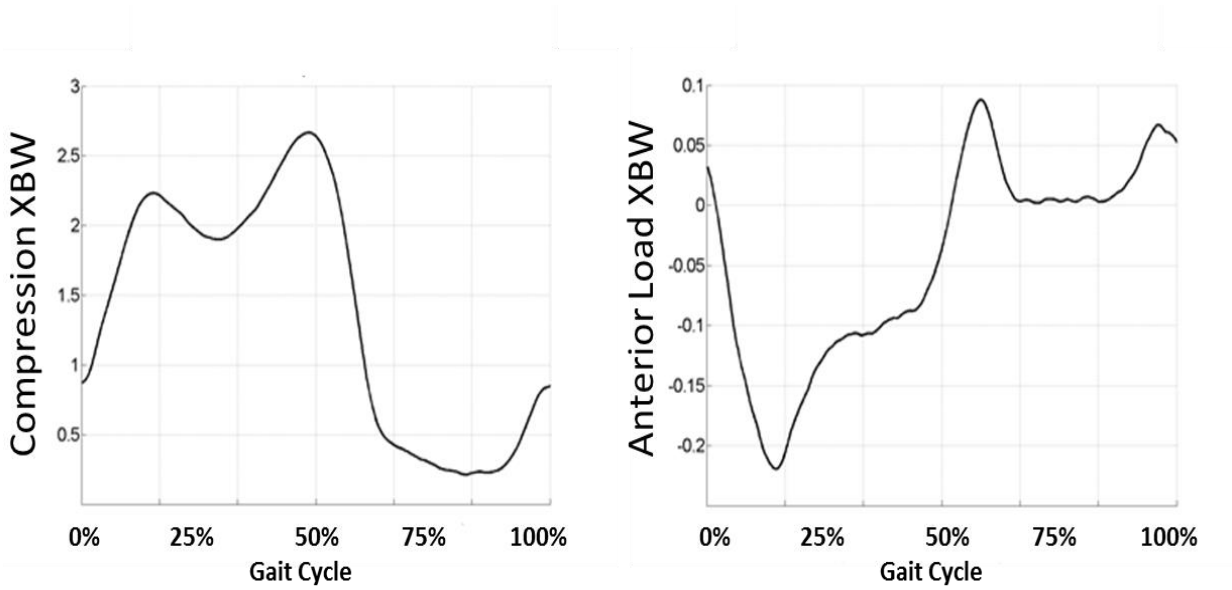


**Figure 4: Compression and anterior loads at the knee during walking obtained from an instrumented tibia data averaged across six subjects [1].**

### 3.3 Results

An increase in hamstrings load increased the anterior load at the tibia (**Error! Reference source not found.** A). A uniform increase in anterior tibia load of approximately 15N, 40N, 90N, and 180N was observed throughout the cycle for hamstring forces of 95N, 147N, 220N, and 257N. The hamstring effect on compressive joint load differed between early flexion and deep flexion. From 20° to 80° an increase in hamstring load caused an increase in compression (**Error! Reference source not found.** B). After 80°, the compressive load decreased with increasing hamstring load. The hamstrings had little effect on the quadriceps load in early flexion but decreased quadriceps load past 80° (**Error! Reference source not found.** C). The posterior ankle force load resulted in minimal changes in A-P load from early to mid-flexion (0-80°). In deep flexion, increasing posterior ankle force load caused a decrease in the anterior tibia load with the largest decrease occurring at 120° (**Error! Reference source not found.** E). In the load range selected, the influence of ankle load on A-P load was an order of magnitude smaller than the effect of the hamstring load. Increasing the posterior ankle force load caused a consistent decrease in joint compressive load across the entire cycle (**Error! Reference source not found.** F). Anterior ankle loads resulted in a large uniform decrease in the quadriceps loads throughout the flexion range (**Error! Reference source not found.** G).

During stance phase of gait, the model was not able to track the posterior loads, although better tracking was achieved using the ankle actuator than the hamstring actuator. The achieved compressive load using the ankle load or the hamstring actuator had small differences from the targeted compressive load (mean 42.95N and 23.43N respectively) (Fig. 5). The hamstring and ankle model configurations predicted similar actuator profiles for the quadriceps and vertical loads.

The hamstrings did not result in consistent changes in medial and lateral lowest point (LP) kinematics (Fig. 6 A,C). The ankle loads caused greater anterior LP translations on both the medial and lateral sides during early flexion (Fig. 6 B,D). During deep flexion, beyond approximately 80°, minimal changes in LP translations occurred for any posterior ankle load.

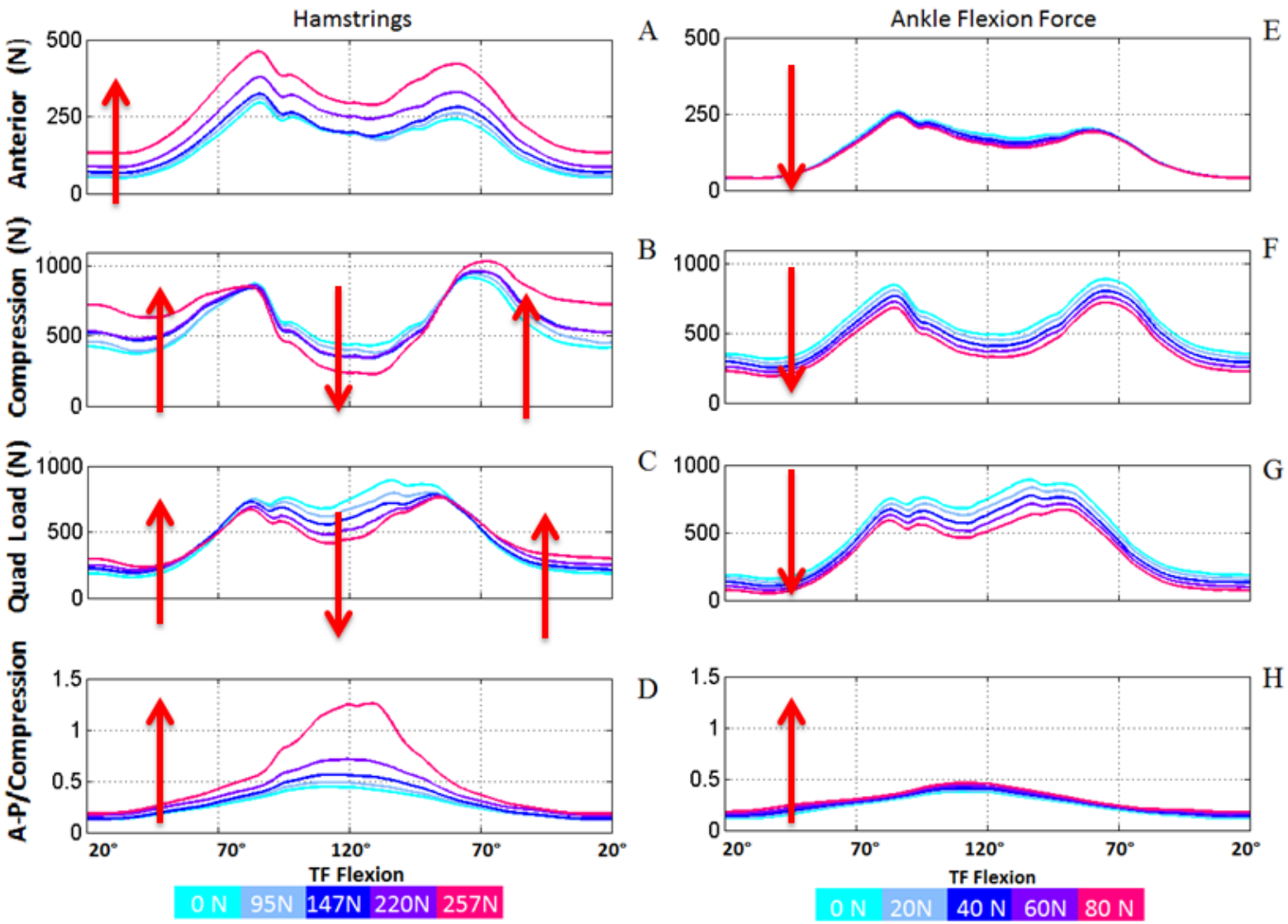


**Figure 5: External loadings effects on anterior knee loads, compression, quadriceps load and AP/compressive ratio. Ankle Load; E-H. Hamstrings: A-D.**

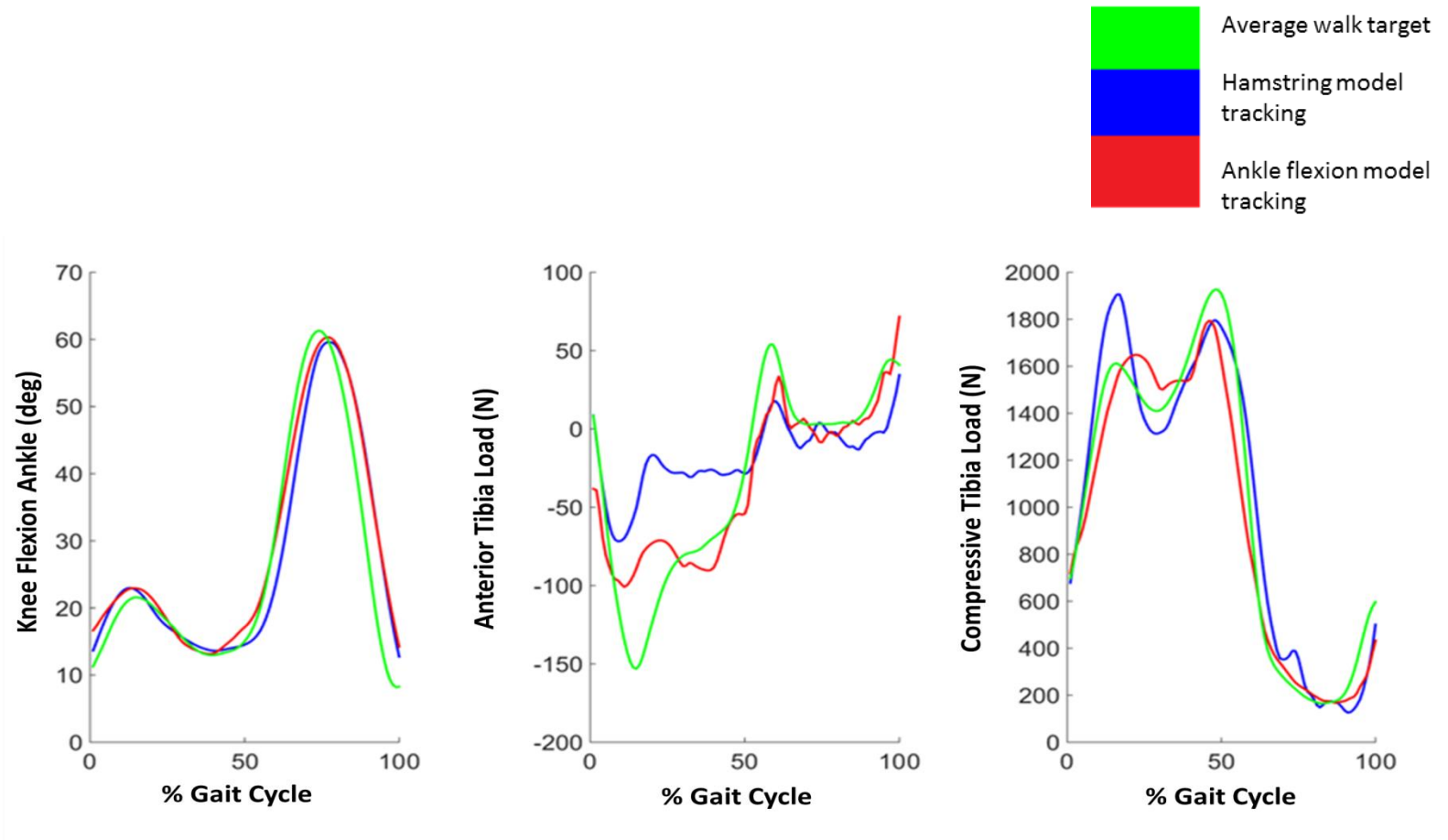


**Figure 6: Model tracking of anterior (left) and compression (right) of the instrumented tibia joint loads.**

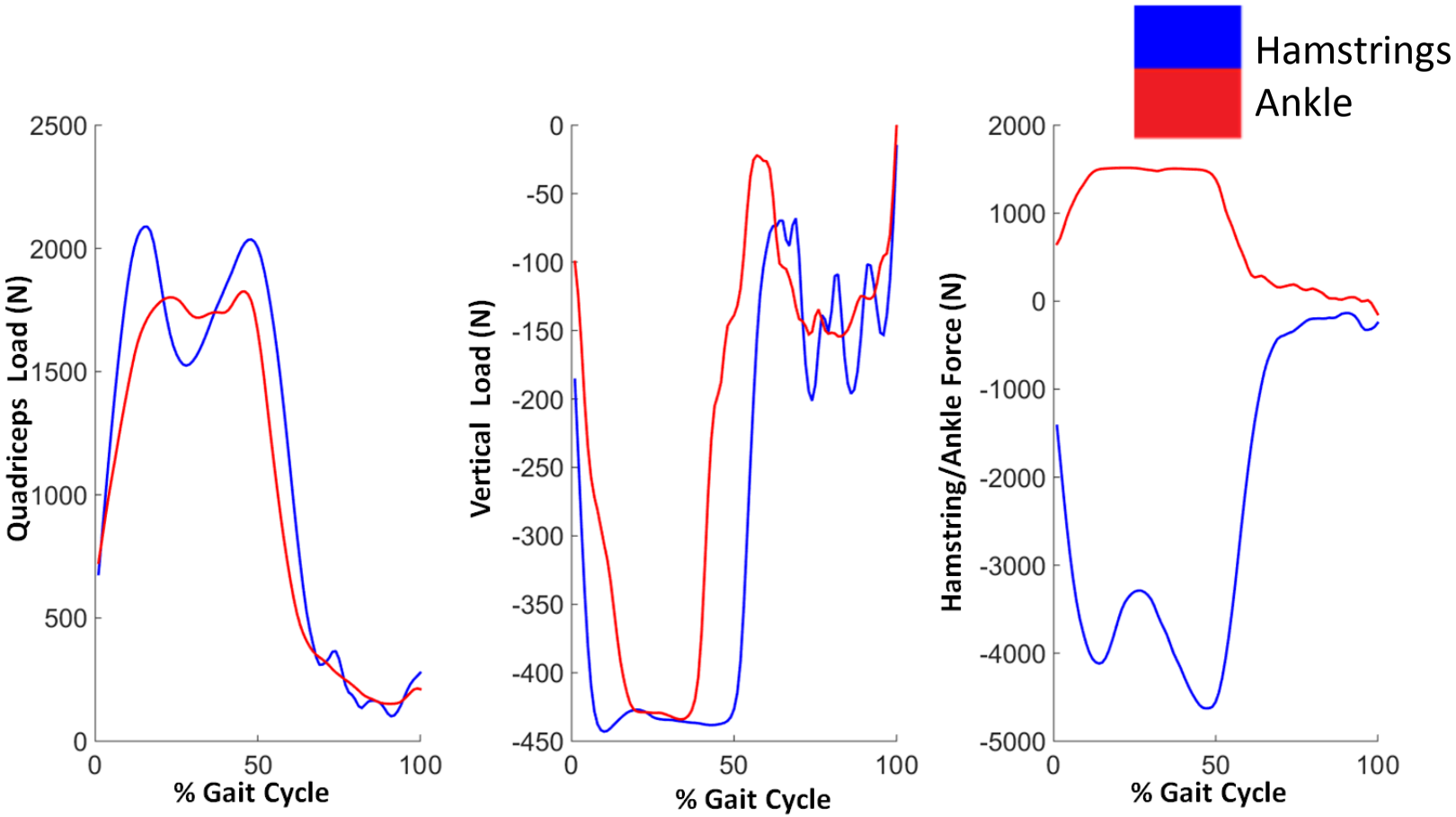


**Figure 7: Model predication of actuator profiles needed to achieve the joint compression and anterior load obtained from the instrumented tibia.**

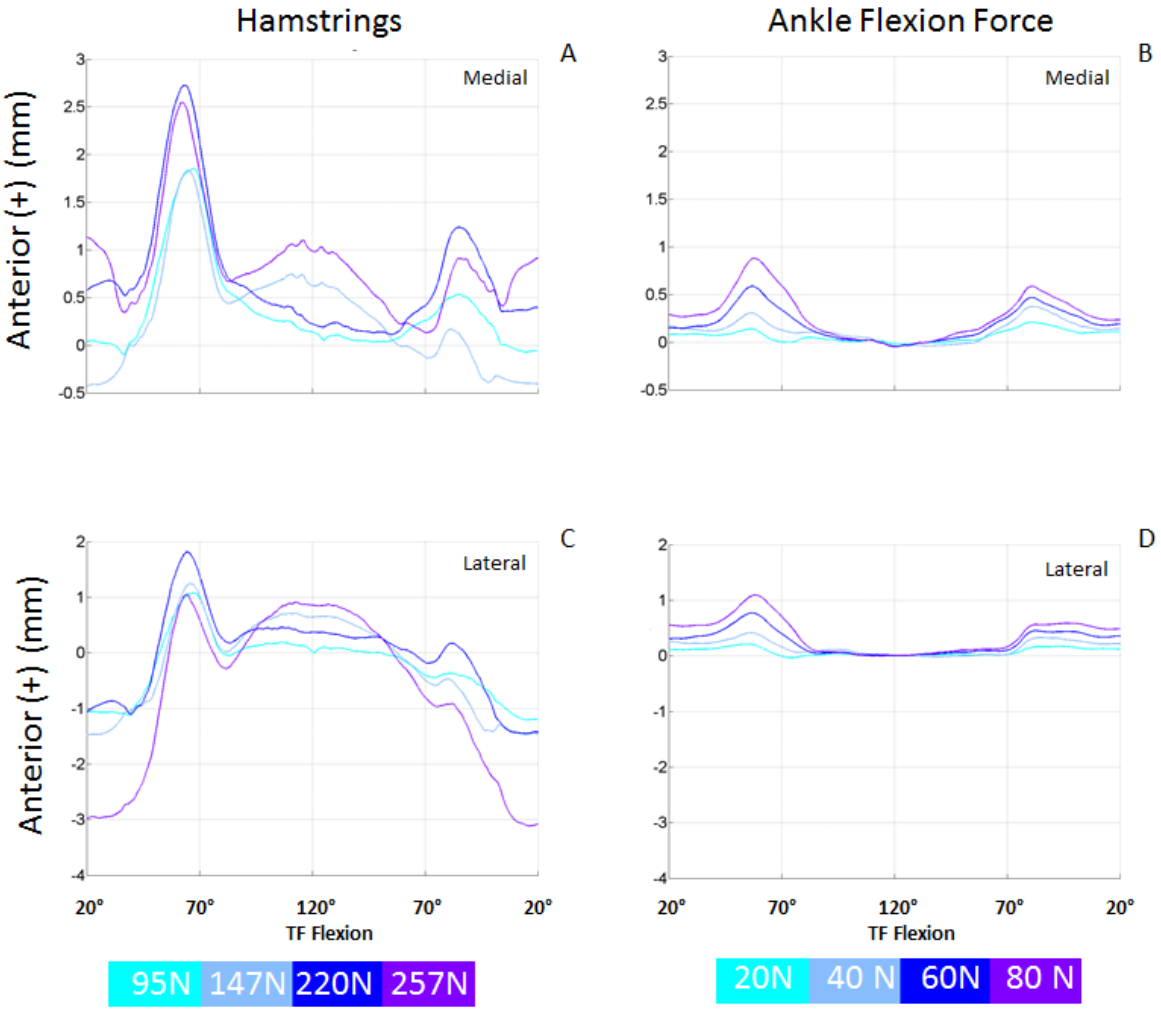


**Figure 8: Lowest point AP translations of the medial and lateral femur. Hamstrings; A, C. Ankle Force B, D.**

## 3.4 Discussion

The goal of this study was to quantify the effects of sagittal plane loads on anterior-posterior (A-P) and compressive tibia forces. This study also aimed to determine the magnitude and direction of hamstrings, quadriceps and ankle loads in a dynamic knee simulator necessary to achieve specific physiological joint loading replicating a gait cycle.

Dynamic knee simulators often use quadriceps loads to control the flexion angle by balancing the flexion moment caused by loads at the hip and ground reaction forces at the ankle. In addition, the loads applied at the hip contribute considerably to joint compressive forces; therefore, a positive association exists between the quadriceps load and compressive joint loads. The contribution of hamstring load to compression and anterior tibia loads are highly dependent on the flexion angle. At early flexion angles, below 90°, the hamstring pulls down on the hip resulting in an increase in tibia compression. At 90°, the hamstring load is perpendicular to the vertical load; therefore, it primarily contributes to A-P load at the tibia. Past 90°, the hamstring likely reduced quadriceps load, which results in a reduction of the compressive load. In contrast, the ankle load had a consistent effect on both compression and quadriceps load throughout the flexion range (Fig. 3). The posterior ankle loads applies an extension moment at the knee joint therefore reducing the loads required by the quadriceps to control the flexion angle. The reduction in quadriceps load will in turn reduce the compressive loads measured at the tibia.

Increasing hamstring loads resulted in a higher anterior tibial forces. D'Lima et al. showed that there is a correlation between the direction of the kinematics and the direction of the tibia load, therefore an increase in anterior tibia was expected as the hamstring load causes anterior translations of the femur relative to the tibia [2]. This differs from the results reported by previous studies which reported that an additional hamstring load caused a decrease in anterior

tibia loads and could even reverse the direction of tibia forces [21-23, 25]. MacWilliams et al. measured tibia loads in a dynamic knee simulator by placing a load cell at the distal tibia. This would mean that the load cell was not only measuring femoral contact forces but also was including the muscle loads applied on the tibia above the load cell. Multiple musculoskeletal models included both the femur contact forces and muscle loads in the reported resultant tibia loads. This explains the differences to the result of the currents study. The joint loads measured in the current study are the result of the femoral contact forces alone similar to the measured loads of instrumented tibiae.

Although the posterior ankle forces, which caused an extension moment at the ankle, drive the tibia in the same direction as the hamstrings, their effect on A-P load was different for the loads used in the experimental set up. The decrease in anterior load was likely a result of the posterior ankle loads decreasing the quadriceps loads and therefore the compressive joint forces. In the current configuration of the KKS, the quadriceps controls the flexion position by adjusting the load to achieve a specific flexion angle. The large effects of ankle load on the quadriceps made it difficult to determine the influence of ankle loads on A-P joint loads since there was no compensation for the loss of compression in the system. The decrease in anterior tibia load was a direct result of the decrease in the quadriceps load. Since the quadriceps are ran in position control a posterior ankle load will only have the ability to decrease the anterior tibia load unless there is compensation for loss in the compression. A posterior tibia load cannot be achieved in this manner, as increasing the posterior ankle load will decrease the quadriceps considerably causing an unloading of the joint. To generate posterior tibia loads the tibia needs to be driven in the anterior direction relative to the femur.

The walk cycle used in the model prediction of loading profiles had high posterior and high compressive loads from heel strike until toe off (Fig. 2). During swing phase, the walk cycle had low compression and minimal anterior tibia loads. An anterior ankle load was predicted to achieve posterior tibia loads while a posterior ankle load was predicted to achieve the anterior tibia forces. This is consistent with resultant ankle loads predicted using inverse dynamics in a gait study [35]. In order to achieve posterior tibia loads using the hamstring configuration, the model predicted that the hamstrings needed to apply an anterior force on the tibia. Physiologically the hamstring can only apply a posterior directed load on the tibia; therefore, it is unlikely that the hamstring muscles can contribute to posteriorly directed tibia loads. The loading profiles generated using the hamstrings and ankle loads is in agreement with the experimental data, in that a loading of the hamstring can only generate an anterior tibia load, while a posterior ankle load can create posterior loads when compensating for changes in quadriceps loads.

Increases in hamstring load did not have a consistent effect on femoral translations. It was also observed that the hamstring configuration did not result in post cam engagement. Multiple studies reported that the inclusion of hamstrings causes anterior femoral translations [21-23]. The hamstrings might have induced an I-E rotation which would mask A-P LP translations. Both posterior ankle load and hamstrings load drives the tibia in the same direction. The posterior ankle loads caused increases in anterior femoral translations until approximately 80º, when it was assumed that post cam engagement had occurred.

Several limitations exist in this study. The study was performed on a TKA knee prostheses mounted to custom fixtures and did not account for the contribution of soft tissue onto the joint load. Cam engagement of the cruciate sacrificing TKA prosthesis used in this study limits A-P translations making it difficult to quantify the effect of applied loads. Loading of soft tissue

structures can alter the loads transmitted to the tibia. Combinations of multiple constant force springs were used to simulate the hamstrings that could have induced internal-external rotation. This change in knee rotation could have influenced the A-P translation of the medial and lateral femoral condyles. The conformity of the tibiofemoral joint also has a large influence on resultant kinetics and kinematics. The data obtained from the instrumented tibia was not the same prosthesis that was used in this study. The differences in conformity between the two prosthetic geometries will affect the magnitude of loads measured at the tibia. Understanding which external loads contribute to these opposing directed tibia loads can provide valuable insight into the mechanics of simulating of dynamic activities in-vitro. Determining the magnitude and direction of applied external loads necessary to create physiological joint loading conditions can influence methodologies used to evaluate new prosthetics design.

## 4. Discussion and Conclusion

In order to obtain physiologically meaningful kinematics from dynamic knee simulations it is essential that consistent physiologically accurate loads are generated at the knee. Multiple factors contribute to a simulator's ability to achieve a specific joint loading condition. The chosen applied external loading conditions, selected musculature, and control systems have a direct effect on the capabilities or range of activities that these systems will be able to simulate. Investigating the individual contribution of externally applied loads on resultant joint forces can help determine if the axes of applied loads are necessary to simulate a dynamic activity.

It was expected that a posterior directed ankle load, which caused anterior translation of the femur relative to the tibia, should result in an increase in anterior tibia forces. The opposite result was observed in this study with small decreases in anterior tibia force and a larges decreases in compression. This counterintuitive result can be attributed to a compensatory decrease in quadriceps loads. A posterior ankle load decreases the flexion moment at the knee, which causes a subsequent decrease in quadriceps load. The decrease in compression is the direct result of the decrease in quadriceps force. The effect of changes in quadriceps loads has a less intuitive effect on A-P joint forces. Increases in quadriceps forces increase load applied at the hip sled, which causes anterior translations of the femur relative to the tibia. Decreases in anterior tibia load were observed during large decreases in quadriceps.

The hamstrings influence tibia motion in the same direction as a posterior ankle load but had a different effect on A-P joint forces for the loads used in the experimental setup. The hamstrings loads caused an increase in anterior tibia loads across the whole flexion cycle.

The second aim of this study was to predict loading profiles for the five axes of the KKS necessary to achieve the joint forces during a simulated walking activity. The hamstring model predicted that the hamstrings needed to push the tibia anterior to achieve the posterior joint forces seen during the early part of the gait cycle. This direction of applied hamstring load is not physiological. An anterior ankle load was predicted to achieve the posterior loads, although the loadcell data showed that posterior ankle load caused decreases in anterior joint forces. This indicates that posterior ankle load can decrease anterior forces because of their influence on the quadriceps force, but a posteriorly directed ankle load does not have the capability to create a posteriorly directed tibia loads. An anterior ankle load resulted in high quadriceps loads and compressive joint forces. The vertical force applied an upward force on the hip to decrease compressive loads at the knee.

Dynamic knee simulators that position the hip directly over the ankle will like have the ability to create high compression and high anterior tibia forces. To expand the range of activities that can be simulated it is necessary to have the ability to generate posterior tibia loads. Hamstring forces only contribute to anterior tibia forces and do not contribute to posterior tibia knee forces. As both the vertical force and quadriceps mechanism cause anterior tibia forces at the knee, it is unnecessary to include a hamstring axis. It is therefore necessary to maintain the ankle load axis, which can apply anterior ankle load to achieve posterior tibia loads. The ankle load axis is highly coupled with the quadriceps mechanism; therefore, changes in the quadriceps and compression will need to be compensated with appropriate forces on the vertical axis.

This study had some limitations including simplifications in the experimental setup and assumptions used in the development of the model. The analog knee used in the experimental setup did not include soft tissue structures that would be present in a cadaveric specimen. Soft

tissue structures are likely to absorb some of the externally applied loads. In addition the type of prosthetic used in the experimental rig will likely effect resultant joint forces. A more conforming articular geometry will result in higher tibia forces.

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# Appendix A

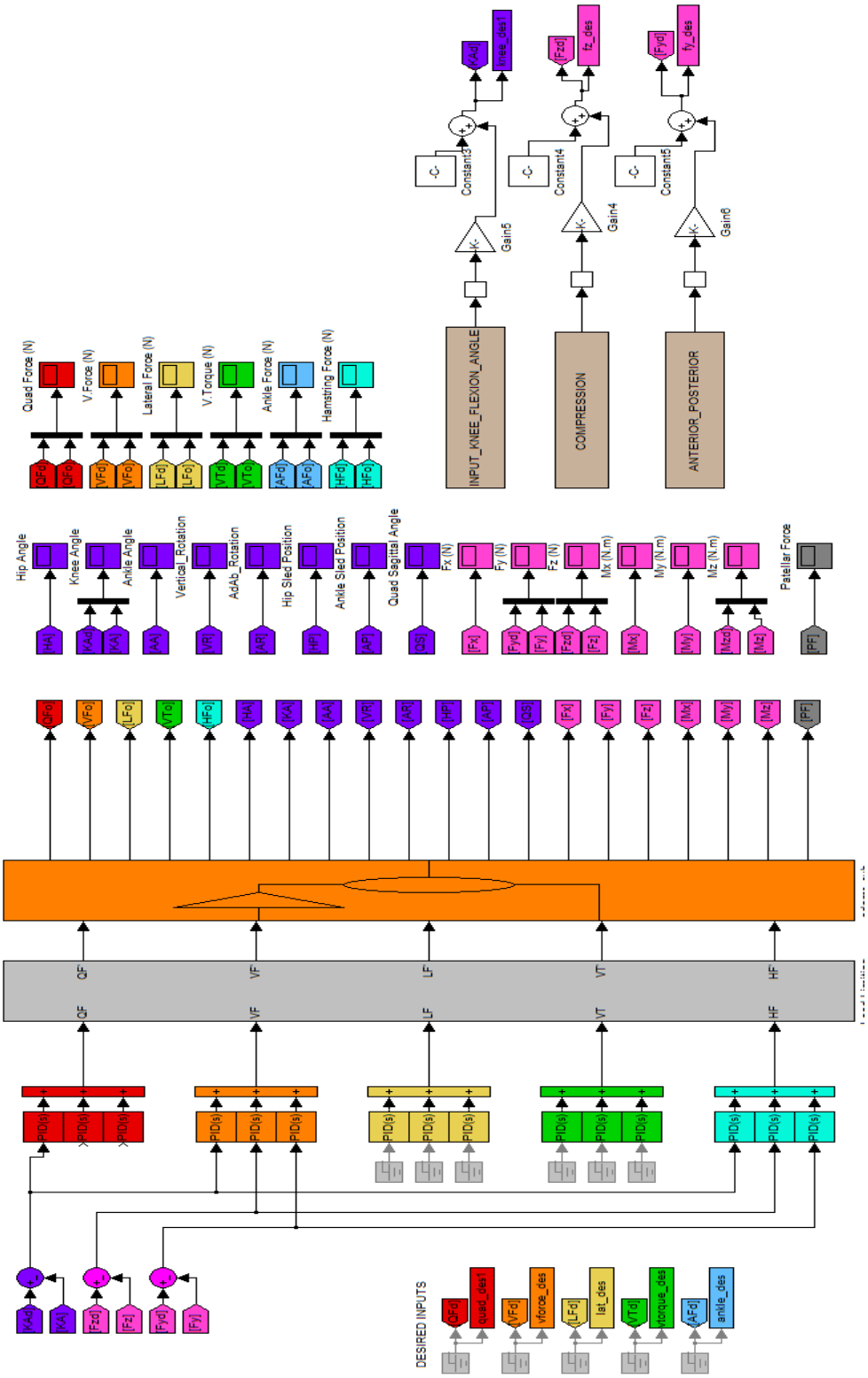


**Figure 9: a diagram of the control system created in Simulink used to create the loading profiles for the KKS model with a hamstring axis included.**