

Effects of Hip Osteoarthritis on Lower Extremity Joint Contact Forces

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ABSTRACT

People with osteoarthritis (OA) suffer from joint degeneration and pain as well as difficulty performing daily activities. Joint contact forces (JCF) are important for understanding individual joint loading, however, these contact force cannot be directly measured without instrumented implants. Musculoskeletal modeling is a tool for estimating JCF without the need for surgery. The results from these models can be very different due to different approaches used in the development of a model that was used for simulation. Therefore, the first purpose of this study was to develop and validate a musculoskeletal model in which lower extremity JCF were calculated at the hip, knee, and ankle in 10 participants with hip OA (H-OA) and 10 healthy control participants using OpenSim 4.0 [simtk.org, 23]. The generic gait2392 model was scaled to participant demographics, then the inverse kinematics (IK) solution and kinetic data were input into the Residual Reduction Algorithm (RRA) to reduce modeling errors. Kinematic solutions from RRA were used in the Computed Muscle Control (CMC) tool to compute muscle forces, then JCF were estimated using the Joint Reaction Analysis tool. Validation included JCF comparisons to published data of similar participant samples during level walking, and movement simulation quality was assessed with residual forces and moments applied at the pelvis, joint reserve actuators, and kinematic tracking errors. The computed JCFs were similar to the overall trends of published JCF results from similar participant samples, however the values of the computed JCFs were anywhere from 0.5 times body weight (BW) to 3BW larger than those in published studies. Simulation quality assessment resulted in low residual forces and moments, and low tracking errors. Most of the reserve actuators were small as well, besides pelvis rotation and hip rotation. The computed JCF were then used in the second portion of this study to determine the effect of group and side on JCF during both the weight acceptance and push-off phases of level walking. It was determined that there was a significant difference in the knee and ankle JCF during the weight acceptance portion of stance phase and at all joints during the push-off phase when comparing the H-OA and control groups on the affected limb. A significant interaction between group and limb was found for the peak hip JCF timing (% stance) during the push-off portion of the stance phase ($p=0.009$). These results demonstrate that H-OA participants experience an earlier peak hip JCF during propulsion on their affected limb. Based on previous research in OA that has examined spatiotemporal measures, this finding suggests that H-OA participants may use step or stride length changes as a strategy to decrease or limit pain and loading on the affected limb. Knowledge of potential JCF differences in H-OA participants, such as timing of the peaks in either portion of the stance phase, could provide useful insight to clinicians and therapists to make decisions on how to proceed with treatment or rehabilitation programs.

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General Audience Abstract

People with osteoarthritis suffer from joint degeneration and pain as well as difficulty performing daily activities, like walking. It is important to understand the forces and loading within individual joints. Musculoskeletal modeling is one way that researchers can estimate these joint contact forces (JCF) without needing a joint replacement implant that can measure these forces. When it comes to modeling simulations, there is a wide variety of results. Therefore, the first purpose of this study was to develop and validate a musculoskeletal model in which JCFs were calculated at the hip, knee, and ankle in 10 participants with hip osteoarthritis and 10 healthy adults. Validation of the model was completed through a comparison between computed results and published data of similar participant samples during level walking. The computed results were similar to the overall trends of published JCF results, however the numerical values themselves were larger than those in published studies. The computed JCFs were then used in the second portion of this study to determine how the two groups and limbs differ during level walking. There was a significant difference in the knee and ankle JCF during the first half of the stance phase and in all joints during the second half of stance when comparing the two groups. The hip osteoarthritis participants also experience an earlier peak hip JCF during the second half of stance phase on their affected limb. This finding suggests that hip osteoarthritis participants may change the way they take a step as a strategy to decrease or limit pain and loading on the affected limb. Knowledge of potential JCF differences, such as timing of the peaks in either portion of the stance phase, could provide useful insight to clinicians and therapists to make decisions on how to proceed with treatment or rehabilitation programs.

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List of Abbreviations

BW	Body Weight
CMC	Computed Muscle Control
EMG	Electromyography
GRF	Ground Reaction Force(s)
H-OA	Hip Osteoarthritis
IK	Inverse Kinematics
K-OA	Knee Osteoarthritis
JCF	Joint Contact Force(s)
OA	Osteoarthritis
RRA	Reduced Residuals Algorithm
THR	Total Hip Replacement

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Introduction

Specific Aims

Osteoarthritis (OA) is the most common form of arthritis and a leading cause of global disability [1, 2]. At least 10% of adults over 60 years of age and more than 20 million Americans are affected by OA [3]. The prevalence of symptoms increases with age, while symptom relief treatment options decrease with age, contributing to OA's harmful impacts [3]. Although exact causes and risk factors are unknown, each case exhibits very similar symptoms in terms of joint degeneration, pain, and difficulty performing daily activities. OA has many physical effects like joint degeneration and pain, swelling, and stiffness, which affect gait and mobility.

There is no concrete agreement on criteria that defines when gait asymmetry is considered pathological [4]. Some researchers believe gait symmetry is having equal values of gait variables on both sides of the body [5]. A certain degree of asymmetry exists in able-bodied gait, as normal walking is believed to be a naturally perturbed state [6, 7]. Natural gait symmetry promotes optimal joint health and ultimately decreases the likelihood of joint degeneration, injury, or discomfort [7].

Participants with hip OA (H-OA) have greater gait asymmetries than healthy controls [4]. According to the World Health Organization, about 80% of participants with OA have restricted movement with 25% unable to perform daily tasks [8]. Participants tend to compensate their movements due to joint pain, causing irregular wear and loading on contralateral and adjacent joints. Participants with end-stage H-OA that compensate for pain begin to walk asymmetrically and are 2.5 times more likely to develop knee OA (K-OA) in the contralateral knee than in the ipsilateral knee [9].

Joint loading during normal activities have contributions from the ground reaction forces (GRF) as well as from the muscles applying forces and moments to the joints [10]. The combination of these loads are known as joint contact forces (JCF). JCF are important because they provide information on joint forces, including stabilization of internal muscle forces, to better understand loading and strain within individual joints. This information cannot be directly measured without an invasive approach, such as an instrumented prosthesis or joint replacement

surgery, however JCF can be estimated with the help of musculoskeletal modeling. By estimating the JCF, clinicians and therapists can make more informed decisions on how to proceed with treatment or rehabilitation programs. In addition, providing this information to clinicians may delay the need for invasive joint replacement surgery.

In this study, musculoskeletal modeling was completed using OpenSim software to estimate the JCF in the lower extremities of participants with H-OA in order to understand the impact that OA has on side-to-side limb differences. The objectives of this study were: 1) to develop and validate a musculoskeletal model to estimate joint contact forces in the lower extremity for participants with H-OA and healthy controls during the stance phase of level walking, and 2) to determine limb differences in JCF between participants with end-stage H-OA and healthy controls during the stance phase of level walking.

Specific Aim 1: Develop and validate a musculoskeletal model (OpenSim) to estimate joint contact forces in the lower extremity for participants with H-OA and healthy controls during the stance phase of level walking.

Specific Aim 2: Determine limb differences in joint contact forces between participants with end-stage H-OA and healthy controls during the stance phase of level walking.

Hypothesis: Side-to-side limb difference will be greater in the H-OA sample when compared to the healthy control group.

Background

Predisposition to OA can be passed through generations, but the pattern in which it is inherited is unknown. In some cases, it is not inherited at all; in fact, it may have been developed after injury or joint overuse. It can also be secondary to other diseases, meaning it can develop when the cartilage has already been damaged by another disease or condition. An additional risk factor for the onset of OA is altered/modified joint loading [11, 12]. Therefore, we must consider body weight as a factor in the development of OA. People who carry more weight on their body exert excess force across lower limb joints when walking or performing daily tasks [13]. Excess

weight or force causes additional stress on articular cartilage, which results in the break-down of the cartilage, causing inflammation and therefore a decrease in the joint space [13].

Severe unilateral H-OA is a disease with effects that commonly extend distally toward the knee [14]. Higher cardiac and ventilatory costs of gait are associated with pain and fatigue from daily activities, and OA reduces the “exchange of potential and kinetic energy” [15]. This, in turn, increases the required muscular work to control movement of the center of mass. The more OA progresses, the greater thru cartilage degeneration in the joint. As degeneration worsens and OA progresses, the patient’s level of pain increases. Participants with H-OA find normal locomotion more tiring than healthy participants because OA changes limb mechanics and disrupts the transfer of potential and kinetic energy [15]. In advanced stages of H-OA, participants are “severely restricted by pain, poor general health, and poor health-related quality of life” [16].

According to the World Health Organization, about 80% of participants with OA have restricted movement and 25% are unable to perform daily tasks [8]. Things like walking, going up and down stairs, standing, sitting, and balancing are all important tasks to study in a H-OA sample as these patients have difficulty completing these activities of daily living [40]. The pain and discomfort experienced by those suffering from H-OA often results in the development of compensatory gait patterns, causing irregular wear and loading on other lower extremity joints. Participants typically compensate for pain by increasing pelvic motion and muscle power generation in the lower limbs [17]. OA at any lower extremity joint leads to lower peak vertical ground reaction forces, extension, and plantarflexion moments, and H-OA limits both knee and hip extension and participants use ankle motion to compensate during the push-off phase of walking [18]. Participants with H-OA never move into hip extension during the stance phase of gait, and experience reduced knee extension [18]. Stance phase of gait is defined as the period from when the foot first touches the ground, heel strike, to when the same foot leaves the ground, toe off. Unilateral OA is typically caused by factors like age, obesity, joint injury or excessive use [11,12,13], while bilateral OA is typically caused by uneven loading on the joint [12], sometimes resulting from altered motor patterns due to prolonged pain. H-OA causes asymmetries in dynamic joint loading at the knee, which could potentially lead to the progression of OA in the contralateral knee [9].

Besides evaluation of gait parameters, treatments are expensive and not always effective. General recommendations for treatment and management of OA include medication, therapy, and surgery [2]. These treatments may help to improve symptoms, but do not delay the progression of OA. The pain and burden of H-OA often culminate in a decision to undergo a total hip replacement (THR). Although THR improves several aspects of gait mechanics, it has not been effective in fully returning patients back to healthy long term gait pattern [19]. Participants also display kinematic adaptations at the ipsilateral ankle and the contralateral hip, which leave them at risk for overloading and joint abnormalities, increasing the probability of joint degeneration [19, 20].

During joint replacement surgery, surgeons can implant an instrumented joint prosthetic that can collect data on the joint mechanics. This can provide information that is otherwise not possible to measure, such as joint contact forces (JCF). Knowledge of accurate JCF allow for determining design requirements for joint prosthesis [21] as well as the ability to provide clinicians and therapists with quantitative data to make decisions on patient treatment and rehabilitation [22].

Although direct measurement is not possible without invasive approaches, practical approaches to calculate an estimation of JCF are available. Musculoskeletal modeling enables researchers to analyze human movement and estimate musculoskeletal loads [23]. Musculoskeletal modeling software systems for biomechanical simulations and analysis have become a popular method for analyzing human movement [24]. These models enable researchers and clinicians to assess biomechanical variables that would otherwise require invasive procedures, such as muscle lengths and muscle forces [24]. JCF are a better predictor of internal loads when calculated using musculoskeletal models because they account for both external forces and muscle forces [25]. Some common options for musculoskeletal modeling software include AnyBody Modeling System [26] and OpenSim [23].

There are a variety of musculoskeletal modeling methods, meaning that one variable could be computed using several different processes or methods. Multiple decisions need to be made in determining how to build the model and calculate the desired output variable. Choices include what software to use, what model to start with, and what process to follow to calculate or estimate the variables of interest. For this research, the variables of interest were the JCF of the

hip, knee, and ankle. The hip, knee, and ankle were all included based on reports that the presence of OA in any joint impacts the other lower extremity joints [9, 18]. H-OA participants exhibit reduced hip extension, flexion, knee extension, and increased ankle motion [18]. Both halves of the stance phase were examined based on findings in a 2015 study on the effect of hip, knee, and ankle OA, which determined that there were significant differences between control and OA groups in GRF during weight acceptance and propulsive phases of stance [18].

OpenSim modeling is an open-source software system that allows users to develop models for movement simulations [23]. OpenSim has been used to study muscle forces and joint reaction forces to estimate variables like JCF in subject-specific models [21, 25, 27, 28]. OpenSim has standard models built for various research purposes. The generic gait2392 model [29] is among the most common and was chosen for this research based on comparisons between other generic models available through OpenSim, as well as AnyBody modeling system that were used in various studies [26, 30, 31, 32]. This is a three-dimensional model of the human body, containing 23 degrees of freedom and 92 musculotendon actuators, representing the 76 muscles in the lower extremities and the torso [29]. Models can be used and modified throughout data processing.

The final choice is the process in which the variable of interest is calculated or estimated. OpenSim has a variety of tools that can be used in variable computations and estimations. To compute JCF, the Joint Reaction Analysis tool was used, but the steps to get here may vary. Some studies use the Residual Reduction tool to minimize modeling and data processing errors, while others choose not to, typically to save processing time or given that residual forces are already acceptable [32]. Some studies use Static Optimization to come up with muscle forces [31-35] while others use Computed Muscle Controls [27]. For this research, the model was scaled to the participant's height and weight, and the inverse kinematic solution, achieved through Visual3d (C-Motion, Bethesda, Maryland, USA), was input into the Reduced Residuals Algorithm (RRA). The results from RRA were used as the input for the Computed Muscle Control (CMC) tool, where the results were used to estimate JCF from the Joint Reaction Analysis [32, 33, 35, 36].

When estimating JCF in OpenSim, studies typically include electromyography (EMG) data. EMG provides information about muscle forces and muscle activation. When combined

with musculoskeletal modeling, researchers are providing the modeling software with information that is otherwise estimated or assumed. The use of EMG provides further accuracy when estimating JCF because it takes away the need to assume anything about the muscle forces. A 2020 study investigating hip JCF, hip muscle forces, and hip co-contraction levels in individuals with mild-to-moderate H-OA found that lower hip JCF were detected in the H-OA group, primarily due to lower hip muscle force production [37]. They were able to evaluate the muscle forces along with the JCF because they also collected EMG data.

To make the necessary assumptions and estimate JCF, the modeling software must have an understanding of the contribution of individual muscle groups. In one case, muscle contributions to hip JCF were calculated based on dynamic optimization solutions for normal walking, solved by Anderson & Pandy, 2001 [38]. Their calculations were used to generate a musculoskeletal model to evaluate relative contributions of muscle forces, gravitational forces, and centrifugal forces to the hip JCF during gait [39]. Bergmann et al. 2001 created a unique database containing hip JCF and gait data from instrumented total hip implants, which was then used as an input for a musculoskeletal model to calculate muscle forces and serve to check validity of calculated results [40, 41]. There is also a public database that stores information regarding the loads acting in human joints, called the OrthoLoad Database [42]. This database includes measurements from instrumented joints that were implanted in patients during joint replacement surgeries.

Musculoskeletal modeling solutions make assumptions regarding the contributions from un-modeled joint structures, like cartilage contact within joints or ligaments, that would likely produce the desired joint kinematics. Because of these assumptions, it is important to test the accuracy of simulations in the context of the specific study [23]. OpenSim JCF simulations have been conducted and compared to *in vivo* measurements and other musculoskeletal modeling calculations for validation [25, 34, 35]. Validating modeling results is often not possible because experimental data that details loading is limited. Several studies have found good agreement with model predictions when compared to experimentally measured data from an *in vivo* database [34, 43].

A variety of methods can be used to validate modeling results. Simply put, validation is the comparison of results between researcher and article publications. One method is to simply

compare solution values to *in vivo* values or values reported in previous literature [37, 39, 42]. For example, this study is calculating peak JCF values in lower extremity joints during walking, so for validation, one could compare the calculated peaks to other calculated peak values, or to *in vivo* values reported in literature. Another method is to compare the general shape or trend of the data [32]. Researchers have reported a wide variety of calculated JCF from modeling because of difference in data collection, availability, and method of processing. Another method for validation is to report residual or RMS error values computed from modeling solutions to show consistency in the model [34, 44].

It has been shown that OA in the lower extremity affects gait parameters, such as GRF, joint angles, and joint moments. In H-OA, one specific model predicted a resultant peak force of 4.3 times body weight (BW) at the hip when walking at 1.36 m/s (4.9 km/h), consistent with the peak force of 4.8 BW walking at 1.39 m/s (5 km/h) measured in another study [39, 45]. However, there is considerable variation in hip contact force measurements reported in the literature, possibly due to walking speed. A recorded mean peak hip contact force of 2.4 BW was reported with participants walking at about 1.08 m/s (3.9 km/h) [40]. Forces of about 5.5 BW at the hip were seen during jogging or fast walking, which occasionally spiking to about 7.2 BW [45].

Motivation, Purpose, and Hypothesis

OA affects a large portion of the population, predominantly older adults, so it is important to understand differences beyond typical gait measurements and understand more about the cause of joint pain and degeneration. When rehabilitation and management of OA are not successful in easing the participant's pain or discomfort, the last option is joint replacement surgery which is expensive and not a long-term solution for pain. Patients do not usually return to a healthy gait pattern after surgery. Performing surgery, especially on an older population, is risky and invasive.

Musculoskeletal modeling is a noninvasive method for researchers to understand both external and internal forces acting on joints. It is important to note that the knowledge of estimated JCF that participants are experiencing could help to inform clinicians and allow them to provide better care and develop better treatment plans. Understanding internal strains and

loads on muscles and joints in the H-OA population could allow clinicians and therapists to make more informed decisions on treatments and rehabilitation programs based on findings in H-OA participants who lack proper gait mechanics based on timing and absolute peak JCF values. If applicable, an assistive device could be recommended to adjust or train gait patterns for OA participants during daily activities, given the information discovered regarding joint contact forces and compensatory patterns in this population. If implemented, an assistive device could provide real-time gait mechanics adaptability by receiving input – such as step time, step length, stance time – from the unaffected limb and informing the affected limb on how and when to step, enforcing a healthier walking pattern in H-OA participants.

Therefore, the purpose of this research was to develop and validate a method for using musculoskeletal modeling to estimate joint contact forces in the lower extremity for participants with H-OA and healthy controls during the stance phase of level walking. The secondary purpose of this research was to determine limb differences in joint contact forces between participants with end-stage H-OA and healthy age-matched controls during the stance phase of level walking. It was hypothesized that side-to-side limb difference will be greater in the H-OA sample when compared to the healthy control group.

Musculoskeletal Model Development & Validation

Abstract

There are a variety of approaches to develop a model when using musculoskeletal modeling software for data processing. There tends to be a lot of variety with reported results as well, which makes validation an important step for any model. The purpose of this study was to develop and validate a musculoskeletal model in which lower extremity joint contact forces (JCF) were calculated at the hip, knee, and ankle in participants with hip osteoarthritis (H-OA) and a healthy control group using OpenSim 4.0 [simtk.org, 23]. The generic gait2392 model was scaled to participant demographics, then the inverse kinematics (IK) solution, kinetic data were input into the Residual Reduction Algorithm (RRA) to reduce modeling errors. Kinematic solutions from RRA were used in the Computed Muscle Control (CMC) tool to compute muscle forces, then JCF were estimated using the Joint Reaction Analysis tool. Validation included JCF comparisons to published data of similar participant samples during level walking, and movement simulation quality was assessed with residual forces and moments applied at the pelvis, joint reserve actuators, and kinematic tracking errors. The computed JCFs were similar to the overall trends of published JCF results from similar participant samples, however the values of the computed JCFs were larger than those published in previous studies [32, 40, 42, 43]. Simulation quality assessment resulted in low residual forces and moments, and low tracking errors. Most of the reserve actuators were small as well, besides pelvis rotation and hip rotation.

Introduction

Most gait parameters – ground reaction forces (GRF), joint angles, spatiotemporal gait variables – can be measured or calculated, however direct measurement of some variables – such as joint contact forces (JCF) – are not possible without invasive approaches or instrumented implants. Inverse dynamics calculate the joint intersegmental force, however the JCF is the actual force applied on the articular surface, accounting for internal loads, internal forces, external forces, muscle forces, and ligament and cartilage contribution. Musculoskeletal

modeling enables researchers to analyze human movement and make estimations based on collected kinematic and kinetic data [24]. Musculoskeletal modeling is unique from other biomechanics analyses because some of its uses include: [29]

- computing maximum isometric forces and joint moments that a muscle generates at any position
- studying how surgical changes in musculoskeletal geometry can affect the moment-generating capacity of different muscles
- generating muscle-driven forward simulations of walking and running to analyze how muscles can contribute to different motions and how joints are loaded

When *in vivo* measurements are not possible, modeling needs to be used to estimate JCF because modeling provides an understanding of human properties – like muscle and tendon properties [46, 47] – that other movement analysis programs cannot provide. Musculoskeletal modeling is a more accurate predictor of JCF because it accounts for both external forces and muscle forces [25]. These models enable researchers and clinicians to assess biomechanical variables that would otherwise require invasive procedures, such as the determination of muscle lengths and muscle forces [24].

OpenSim modeling is an open-source software system that allows users to develop models for movement simulations [simtk.org, 23]. The generic gait2392 model [29] is among the most commonly used model for lower limb walking studies and was chosen for this research based on comparisons between other generic models available through OpenSim and the AnyBody modeling system [26, 30, 31, 32]. The gait2392 model was chosen over the gait2354 model, both readily available OpenSim models, because it includes more muscles than the gait2354 model [29]. This choice allowed for calculation of more muscle forces and thus more muscles to drive the simulation and JCF calculation. This choice sacrificed processing time, as the number of muscles was reduced from the gait 2354 model to the gait2392 model by Anderson in order to improve simulation speed for demonstration purposes [48]. The gait2392 model does not include the arms on the model. The lack of arms could impact the model results given the importance of arm swing to overall gait stability [60].

In OpenSim, JCFs can be computed using the Joint Reaction Analysis tool using the muscle recruitment solution, movement kinematics and external forces as inputs. Some studies use the Reduced Residuals Algorithm (RRA) tool which uses a tracking controller to follow the movement kinematics and external loads inputs, while others choose not to, typically to save processing time or given that residual forces are already acceptable [32]. Some studies use a static optimization approach to determine the muscle recruitment solution [31-35] while others use a Computed Muscle Control (CMC) algorithm [27] which compute muscle excitation levels that drive the model towards the desired kinematic trajectory.

RRA is intended for movements like walking and running and therefore was used in this study to reduce errors due to modeling assumptions. These errors can include noise, and motion capture errors that can lead to dynamic inconsistencies [23]. However, there is some floating or leaning in the model due to inaccuracies in mass distribution and geometry of the torso. For this reason, the torso center of mass was adjusted during RRA. If the minimum/maximum allowed residual values are too restrictive or too lenient, the RRA results cannot generate realistic muscle function from CMC. CMC is a great tool to determine muscle excitations and muscle forces that drive the model to match the desired kinematics and was used in this study. However, the initial values of the muscle states (generalized coordinates and speeds) are unknown for the first 0.030 seconds, thus it is advised to start CMC at least 0.030 seconds before the time interval of interest [23]. Muscle forces determined from CMC are used to determine JCFs from the Joint Reaction Analysis tool.

A variety of methods can be used to validate modeling results. Simply put, validation is done to confirm the accuracy of the research being done. In this study, the research is being validated by comparing results between experimental and published data/results. This study uses JCF values that were calculated based on experimental data collection and compares these results to *in vivo* measurements obtained from instrumented implants [40, 42, 43] or modeling-based calculation results reported in previous literature [32]. For example, this study is calculating peak JCF values in lower extremity joints during walking, so for validation, one could compare the calculated peaks to other calculated peak values, or to *in vivo* values reported in literature. Another method is to compare the general shape or trend of the data [32]. Researchers have reported a wide variety of calculated JCF from modeling, typically ranging from 1.5 to 4 times

participant's body weight during level walking (walking speed ranging from about 0.87 to 1.4 m/s) [27, 31, 32, 35]. Due to differences in data collection (inclusion criteria, participant walking speed, equipment usage), availability of data (EMG or MRI data collection), and processing method (software usage). Simulation quality is also important to assess in terms of dynamic consistency in the modeling solution and to determine how well the muscles are driving the movement simulation [34, 44].

The purpose of this research was to develop and validate a method of musculoskeletal modeling in which lower extremity JCF were calculated at the hip, knee, and ankle in a hip osteoarthritis (H-OA) and healthy control sample using OpenSim 4.0 [23]. Participants included in this study were asked to walk at a self-selected pace along a 40-m walkway while kinetic and kinematic data were collected [49]. Metrics for validation included comparison of JCF trend and values to published modeling calculation of JCFs and *in vivo* JCF measurements from instrumented implants in similar participant samples. Metrics for simulation quality included residual forces and moments, RMS error values, and reserve actuators.

Materials and Methods

Model Development: A generalized 3D musculoskeletal model of the legs and torso in OpenSim 4.0 [simtk.org, 23] with 23 degrees of freedom and 92 Hill-type musculotendon actuators [29] was used to develop walking simulations (gait2392). Musculotendon actuators were governed by the force-length-velocity relation, and both passive and in-parallel contractive elements contributed to the muscle force [47]. The musculoskeletal model was scaled using the Scale Model Tool, according to participant demographics (height and mass) obtained during the data collection session. Motion capture (Motion Analysis Inc, Santa Rose, CA) and force plate (ATMI Inc, Watertown, MA) data from a previously collected walking study [49] were used for simulation development. Experimental kinematic and kinetic input data were filtered at 6 Hz in all OpenSim tools used [44]. The inverse kinematics (IK) solution, achieved using a global least squares optimization method in Visual3D (C-Motion, Bethesda, Maryland, USA) [50], and ground reaction forces (GRF) were input in the RRA Tool to minimize errors due to modeling assumptions and experimental data collection to improve dynamic consistency [23, 51]. The model was updated to incorporate the suggested mass and torso center of mass location

adjustments. The inverse kinematics solution was also adjusted to minimize these errors. The adjusted model and walking kinematics from RRA were then input in the CMC Tool to determine the muscle excitations and associated muscle forces that drove the model to track the desired kinematics from RRA [23]. The CMC solutions were then analyzed using the Joint Reaction Analysis Tool to compute lower limb JCF [32, 33, 35, 36]. Data processing workflow in OpenSim follows Figure 1.1.

Figure 1.1: Data Processing Workflow in OpenSim

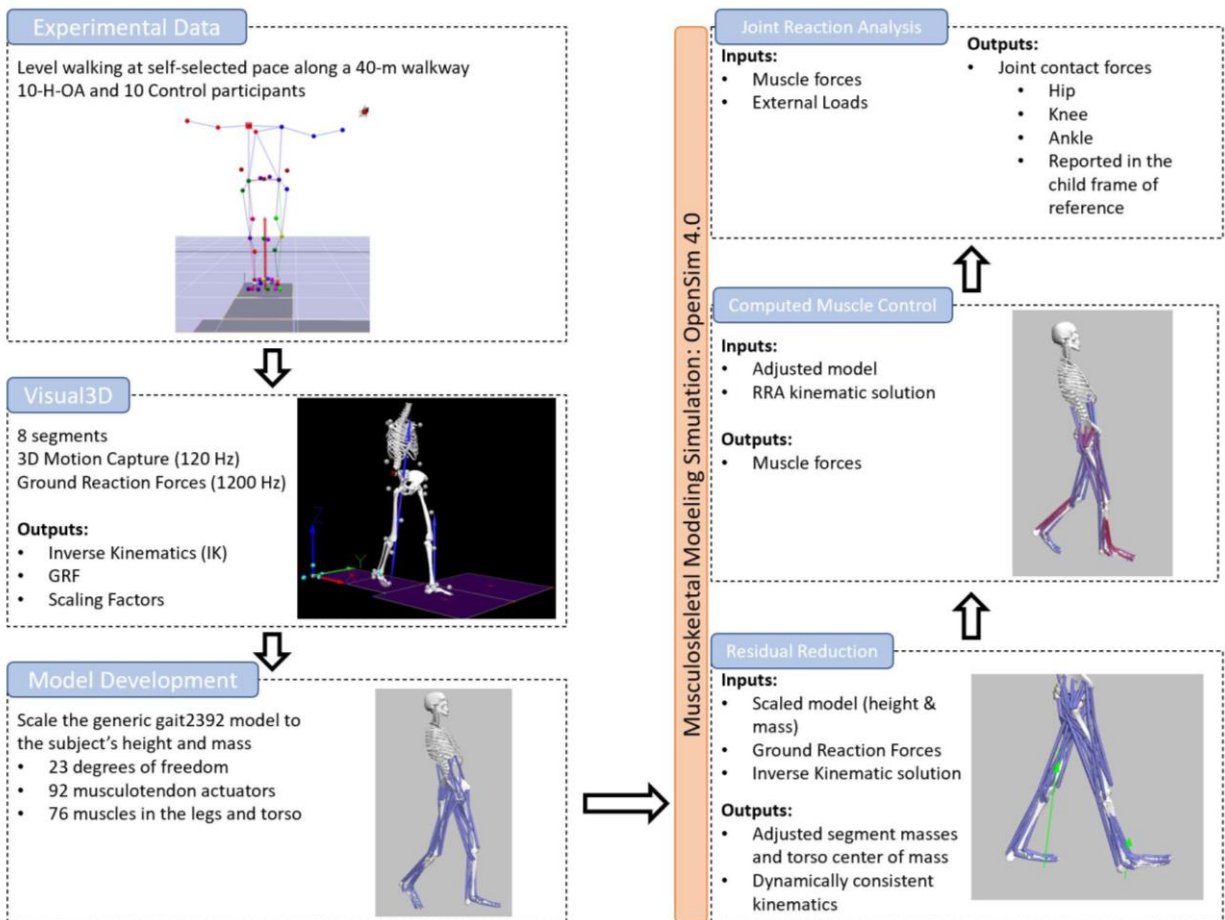


Figure 1.1: Workflow used for processing experimental data, model development, and generating walking simulations. In a previous study, level walking data were collected and processing in Visual3d. Motion capture and GRF data were sampled at 120 Hz and 1200Hz, respectively. IK, GRF, and scaling factors were exported from Visual3d for model development and processing in OpenSim 4.0 [23].

Movement Simulation Quality: Residual forces and moments from the movement simulations (output from CMC) were normalized by participant body weight (BW) for forces (FX, FY, FZ) and BW*height for moments (MX, MY, MZ), so these values are unitless. For good simulation quality, residuals should be low to ensure dynamic consistency. Joint reserve actuators from CMC solutions were normalized by participant BW, so these have units of (m). The reserve actuators are torques added to each degree of freedom to enable the simulation to run [23]. If joint reserve actuators are high, the muscles may not be effectively recreating the net joint moment. Tracking error in the kinematic solutions was calculated for each trial using the equation below (eq. 1.1). This equation is determining the RMS error between the CMC solutions from OpenSim and the IK solution from Visual3D. The difference between these two solutions were found at each time point within a trial, squared, divided by the number of time points in the trial (n), then the square root was taken. RMS Error values for each trial were averaged across participant, then across group for a single value per group, per variable. This was done for each limb independently. If tracking errors are high, the simulation is not effectively tracking the experimental data.

$$RMS\ Error = \sqrt{\frac{(CMC\ solution - IK\ solution)^2}{n}}, \text{ n= number of time points across trial} \quad (\text{eq. 1.1})$$

Joint Contact Force Validation: While direct validation was not possible with this study due to challenges collecting JCF *in vivo*, peak values and overall shape of joint contact forces were compared to prior literature. JCF calculations across stance phase were averaged across participants. The results of this study's H-OA participants affected limb were compared to both prior musculoskeletal modeling calculated JCF [32] as well as *in vivo* JCF measures reported in published literature [40, 43] and obtained from the OrthoLoad database [42] for similar participant samples. In Hoang et al. 2019, 18 H-OA participants (age = 64.0 ± 7.0 yr) were asked to walk barefoot at a self-selected pace (walking speed = 1.17 ± 0.16 m/s) along a 10m walkway while outfitted with retroreflective markers [32]. The musculoskeletal modeling calculations from Hoang et al., 2019 were chosen for comparison with this study's calculated results (Fig. 1.2), as calculations on models both with and without electromyography (EMG) data were included [32]. Bergmann et al. 2001 included 4 patients with instrumented hip implants (age =

51-76 yr) who were asked to perform nine different daily activities. The normal walking trials (walking speed = 1.09 m/s) were used for this comparison [40]. Damm et al. 2017 included participants with instrumented hip (age = 56 yr) and knee (age = 71 yr) implants who were asked to perform walking trials in a gym at a self-selected speed (walking speed = 0.83-1.1 m/s) 12 months postoperatively [43]. Implant data for the hip (age = 62 yr, 12 months postoperative) and knee (age = 62 yr, 20 months postoperative) were available in the OrthoLoad database, so these were the only two joints presented for this comparison.

Results

In this study, 10 H-OA participants (age = 63.9 ± 6.67 yr) were asked to walk barefoot at a self-selected pace (walking speed = 1.27 ± 0.05 m/s) along a 40m walkway while outfitted with retroreflective markers [49]. In all three joints, the JCF values found in this study were larger than those found by Hoang et al. Of these three joints, the second peak of the ankle shares the closest resemblance to the calculated results from the chosen study for comparison (Fig. 1.2c) [32]. The shape profile of all curves are similar, as there are two clear peaks, however the ankle JCF calculated in this study had a peak in the first half of the stance phase, unlike Hoang et al.'s EMG-assisted and static optimization curves (Fig. 1.2c).

In vivo measurements for the hip and knee from two studies [40, 43] and the OrthoLoad database [42] are reported for comparison to this study's calculated results (Fig. 1.3). In both joints, the general shape of all curves were similar with two clear peaks, however the JCF results of this study were anywhere from 0.5BW to 3BW larger than those of the *in vivo* measurements.

Movement simulations had low residual forces and moments throughout the stance phase (Fig. 1.4) as well as low peak values (Table 1.1). Both groups and limbs followed the same general pattern, besides the vertical moment (MY). Overall, tracking errors between the IK and CMC solutions were low, with the largest RMS errors in pelvis rotation and hip rotation (Table 1.2).

Reserve actuators throughout the stance phase are represented in Figure 1.5. Most reserve actuators remain near zero with large peaks occurring predominantly among the control participants. These peaks occurred at either the beginning of the stance phase in hip flexion, knee

angle, and ankle angle, or around the same time as the peaks in GRF in hip adduction and rotation.

Figure 1.2: Comparison of lower limb JCF to prior model-estimated JCF

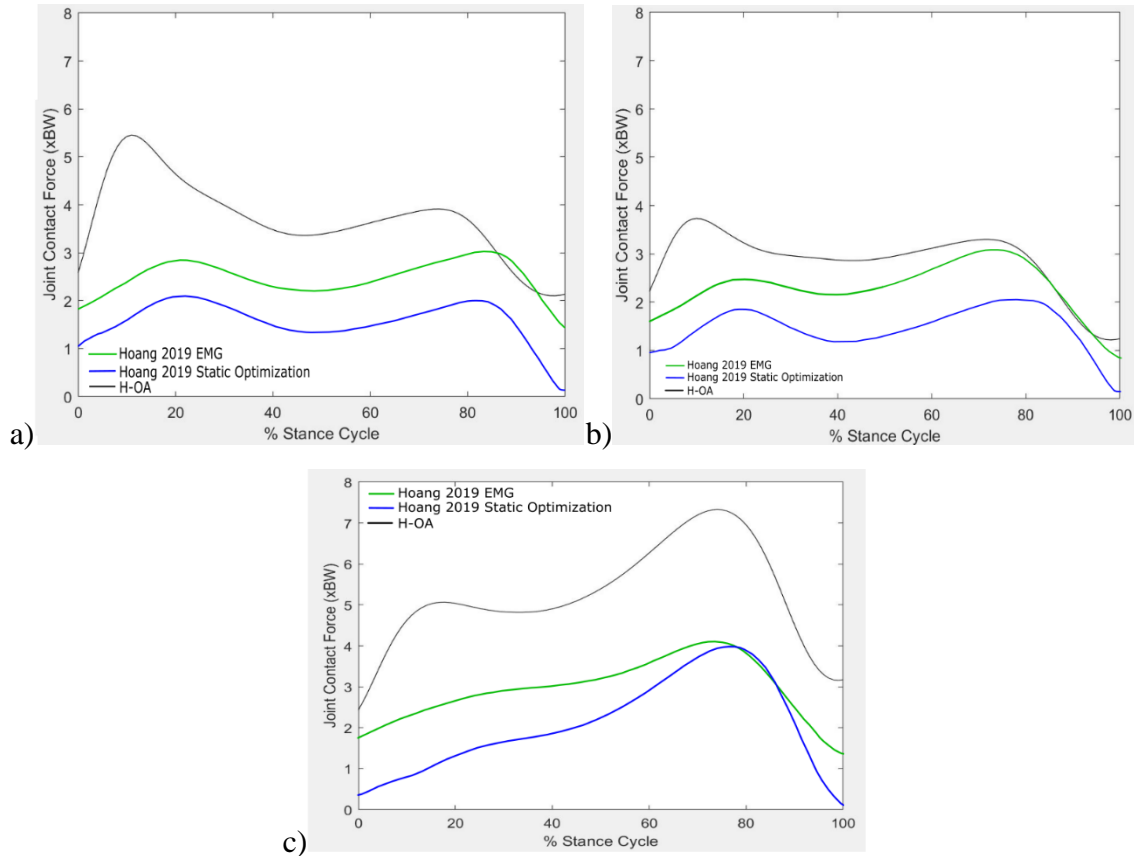


Figure 1.2: These plots contain this study's calculated H-OA JCF values from the affected limb averaged across participants (black) and trend lines of data extracted from Hoang et al. 2019 [32] including an EMG-assisted model (green) and a static optimization model (blue) for comparison of model-based calculations. All joints follow the same overall shape as the Hoang et al. 2019 curve. a) Hip JCF over the stance phase; b) Knee JCF over the stance phase; c) Ankle JCF over the stance phase; The ankle JCF plot (c) shows the closest resemblance between the other modeling calculations.

Figure 1.3: Comparison of simulation results to *in vivo* measurement

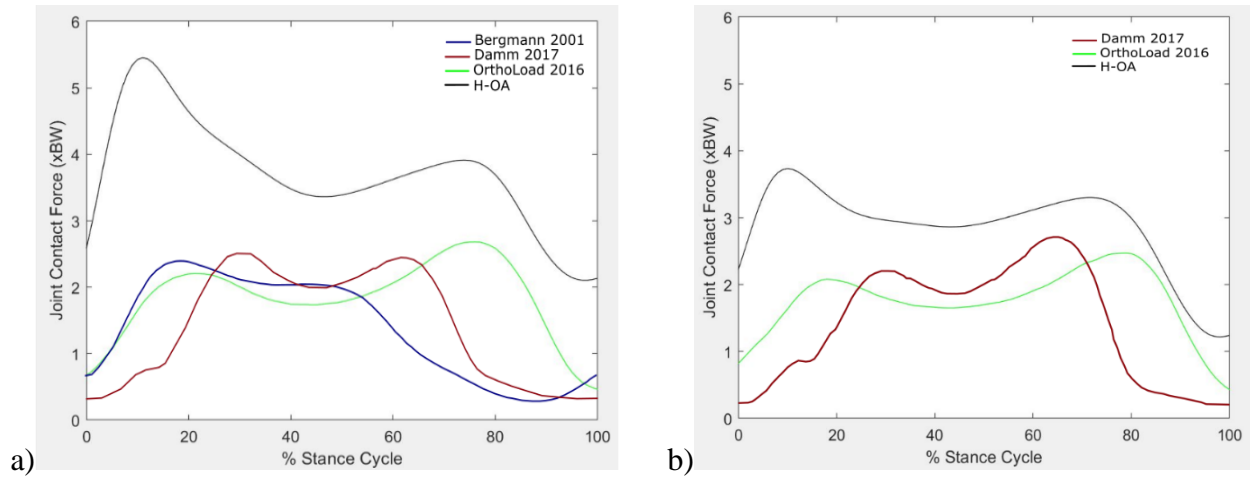


Figure 1.3: These plots contain this study's calculated H-OA JCF values from the affected limb averaged across participants (black) and trend lines of *in vivo* data extracted from Bergmann et al. 2001 (blue) and Damm et al. 2017 (red) [40, 43] as well as data from the OrthoLoad database (green) [42] for comparison to *in vivo* measurements. The H-OA curve follows the overall shape of the *in vivo* curves, although with larger values. a) Hip JCF over the stance phase; b) Knee JCF over the stance phase.

Figure 1.4: Residuals from CMC solutions

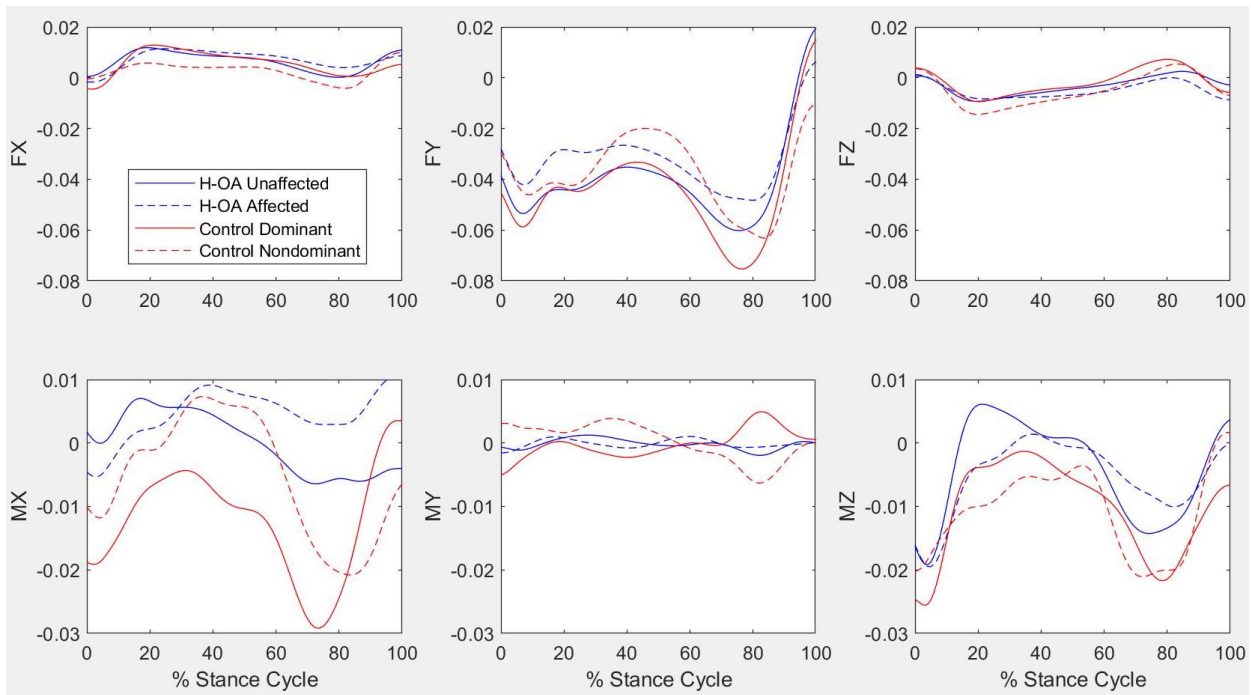


Figure 1.4: Residuals were pulled from the CMC solutions. Forces were normalized by participant's body weight. Moments were normalized by participant's body weight and height. All forces and moments residuals are low. Both groups and limbs seem to follow the same trends besides the vertical moment (MY). Vertical force (FY) has the largest residuals.

Table 1.1: Peak Residuals (unitless) from CMC solutions

	H-OA		Control	
	Unaffected	Affected	Dominant	Nondominant
FX	0.0119	0.0114	0.0129	0.0100
FY	0.0602	0.00483	0.0754	0.0632
FZ	0.0093	0.0087	0.0094	0.0144
MX	0.0070	0.0106	0.0292	0.0208
MY	0.0020	0.0016	0.0050	0.0063
MZ	0.0192	0.0195	0.0256	0.0211

Table 1.1: Absolute peak residuals pulled from CMC solutions. Forces were normalized by participant's body weight. Moments were normalized by participant's body weight and height. All peak residuals look great and are close zero. Residual peaks were accepted if they were less than ~ 0.02 .

Figure 1.5: Reserve Actuators (m) from CMC solutions

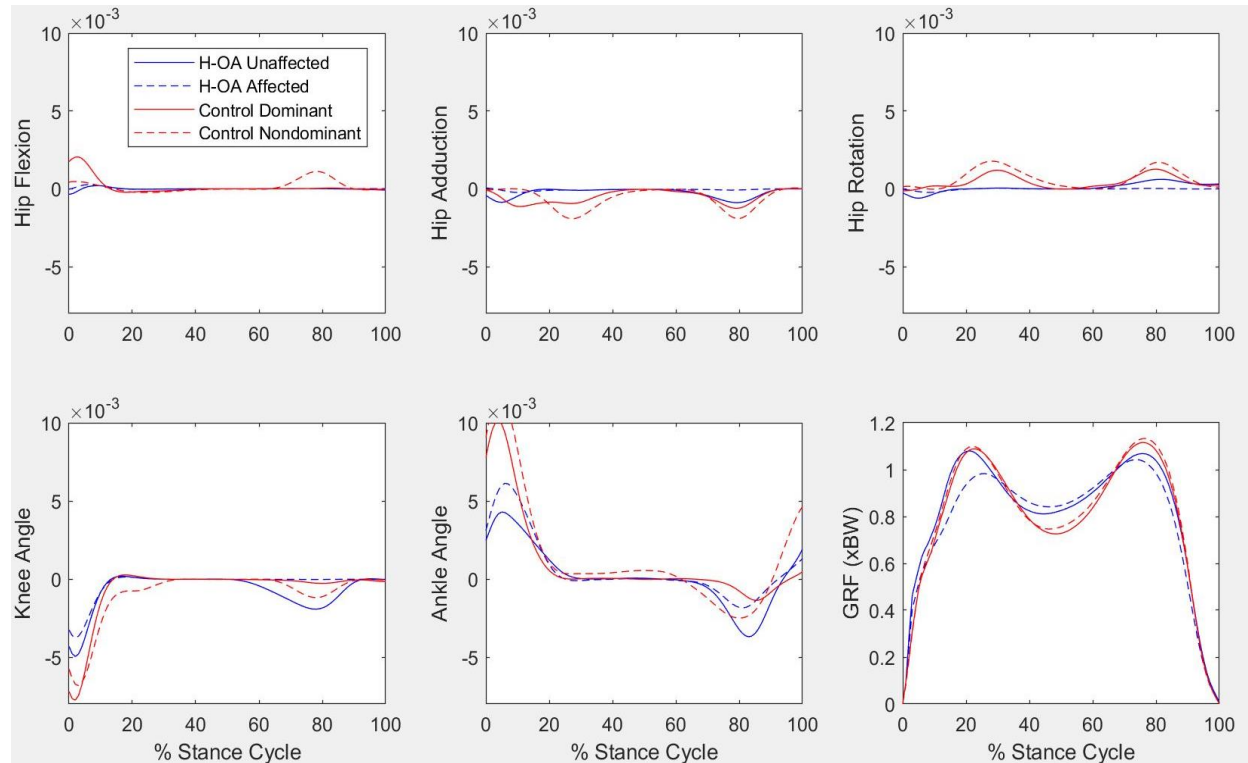


Figure 1.5: Reserve actuators were pulled from the CMC solutions, normalized by participant's body weight. Reserves remain relatively close to zero, with some spikes occurring, particularly in the control group. These spikes seem to occur either at the very beginning of the trial (as in hip flexion, knee angle, and ankle angle) or at the same time as the first and second GRF peaks (as in hip adduction and hip rotation)

Table 1.2: RMS Error Values

	H-OA		Control	
	Unaffected	Affected	Dominant	Nondominant
Pelvis Tilt (°)	1.25 ± 0.88	1.72 ± 1.23	2.37 ± 1.47	1.84 ± 1.15
Pelvis List (°)	11.1 ± 8.81	16.5 ± 14.4	12.9 ± 7.03	12.1 ± 7.99
Pelvis Rotation (°)	28.1 ± 19.6	35.2 ± 22.9	53.4 ± 24.2	47.4 ± 21.2
Pelvis TX (cm)	5.14 ± 4.64	7.57 ± 6.27	7.16 ± 3.13	6.15 ± 3.18
Pelvis TY (cm)	6.84 ± 5.75	8.49 ± 6.27	7.98 ± 5.15	9.44 ± 6.44
Pelvis TZ (cm)	1.54 ± 1.00	2.20 ± 1.67	3.10 ± 1.17	2.99 ± 1.25
Hip Flexion (°)	6.49 ± 4.14	8.31 ± 5.38	10.3 ± 4.79	9.31 ± 5.99
Hip Adduction (°)	4.27 ± 4.39	4.99 ± 3.88	5.18 ± 2.32	5.29 ± 2.89
Hip Rotation (°)	26.6 ± 25.9	23.0 ± 12.4	17.7 ± 8.58	52.1 ± 20.9
Knee Angle (°)	4.80 ± 7.21	2.91 ± 3.54	0.36 ± 0.12	7.05 ± 2.52
Ankle Angle (°)	4.17 ± 3.81	4.93 ± 5.11	7.85 ± 7.77	4.83 ± 1.89
Lumbar Extension (°)	3.35 ± 2.93	3.15 ± 2.49	6.02 ± 4.26	5.52 ± 3.69
Lumbar Bending (°)	1.82 ± 1.63	1.96 ± 1.37	3.43 ± 1.50	3.74 ± 1.59
Lumbar Rotation (°)	2.00 ± 1.76	2.54 ± 2.01	4.06 ± 2.75	3.95 ± 2.81

Table 1.2: Tracking error values are reported as mean ± standard deviation. RMS error was calculated between the CMC solutions and the IK solutions over stance phase. This includes all trials that met inclusion criteria (between 3 to 7 walking trials per subject) for all participants (n=20) for a total of 82 trials. Most tracking errors are low (joint angles < 5°, pelvis translations < 10cm); however, pelvis rotation and hip rotation have large RMS error values.

Discussion

All the published studies included in comparison of this study's results included participants from similar populations (participants with H-OA) or participants with implants that allowed for direct measurement at the joint of interest. All participants were close in age, ranging from 51-76 years of age with participants from this study were in the middle of this range at 63.9 ± 6.67 years of age. In all cases, participants were asked to walk at a self-selected pace, which ranged from 0.83-1.17 m/s while this study's participants walked a little faster at 1.23 ± 0.05 m/s.

Overall, the general shape of the JCFs during the stance phase of gait was similar between the data used in this study and the data collected and published in previous studies. In each comparison plot (Fig. 1.2-1.3), there are two clear peaks in the weight acceptance and push-off phases when the foot was contacting the ground. The H-OA calculated ankle JCF curve had a prominent peak in the weight acceptance phase, unlike that illustrated in the EMG-assisted and static optimization curves (Fig. 1.2c). This could be due to many factors – Hoang et al. processed marker trajectories, GRF, and EMG signals in MATLAB while this study processed marker trajectories and GRF in Visual3D; Hoang et al. did not implement the RRA tool while this study did [32]. The *in vivo* curves pulled from publications [40, 43] had a narrower stance phase than H-OA and the data present in the OrthoLoad database [42] (Fig. 1.3). This could be attributed to different data collection methods, participant walking speeds, or possible inclusion of some points before and after heel strike and toe off. One limitation when comparing experimental data for preoperative participants with implant measurements is that these implant participants are post-operative. Preoperative participants are likely experiencing pain that post-operative participants may not be. Participants from the OrthoLoad database have undergone a procedure and received an implant, so there may be differences – walking kinematics, joint function, muscle behavior, walking speed – between the groups that cannot be accounted for.

Regarding comparison of JCF values, the closest similarity was between the knee JCFs from this study and the EMG-assisted model from Hoang et al. [32] (Fig. 1.2b). The second peak in the H-OA curve is about 0.25BW larger than the EMG-assisted curve. Modeling simulations tend to overestimate when calculating JCF. This can be seen by comparing the model-based JCFs extracted from Hoang et al. (Fig. 1.2) and the implant measurements from the OrthoLoad database and data extracted from Bergmann et al. and Damm et al. (Fig. 1.3). Looking at the knee in particular, the calculated JCFs are about 0.5BW larger throughout stance phase than those measured using the instrumented implant.

Confidence in the walking simulation outputs is increased because of low residuals and tracking errors. The residuals from the CMC solutions remain approximately zero for the entirety of the stance phase (Fig. 1.4). Vertical force (FY) had the largest residuals and H-OA unaffected limb, control dominant limb, and control nondominant limb were outside of the accepted range of < 0.02 (peaks = 0.0602, 0.0754, 0.0632 respectively, Table 1.1). The reserve actuators from

the CMC solutions remain approximately zero with some larger spikes occurring. Looking at hip flexion, knee angle, and ankle angle, these spikes occur at the very beginning of stance phase (Fig. 1.5). In hip adduction and hip rotation, these spikes aligning closely with the weight acceptance and propulsion GRF peaks (Fig. 1.5). The reserves activate when the musculotendon actuators cannot produce the necessary amount of force at a given point in the simulation. Given that overall reserve actuators are low indicates that the muscles in the model were able to reproduce the experimentally observed walking pattern.

Muscle weakness on the H-OA side and in H-OA participants when compared to control participants was reported in the hip adductors and abductors, and hip and knee flexors and extensors [52]. These muscle groups were compared between limb and group. Hip flexors, extensors, adductors, quadriceps, and hamstrings show lower muscle forces in the H-OA group compared to the control group, and in the affected limb compared to the unaffected limb (see Appendix). A limitation to this study could be that the potential muscle weakness was not addressed in the model. This may have affected the JCF calculations and should be considered in future studies.

The RMS error serves as a representation of the difference between values predicted and observed, or between the CMC and IK solutions. RMS error should be small to ensure that the movement simulation is an adequate reproduction of the experimentally observed movement. Most of the hip, knee, and ankle joint angles were less than $\sim 5^\circ$ and pelvis translations less than 10 cm. This study had low RMS error values for all joints besides pelvis rotation and hip rotation, which had the largest RMS errors (Table 1.2). This could indicate errors in marker placement for markers around the pelvis and hip for some participants. Understanding this limitation helps with the interpretation of JCF results at the hip, as the hip seems to have much higher JCF values relative to published results (2.0-3.0BW larger) versus the knee which is only slightly higher than published results (0.25-1.0BW larger) (Fig. 1.3). Across both groups and limbs, the tracking errors remained consistent.

A potential limitation to this study is how well the OpenSim gait2392 model did or did not represent a H-OA population. The model is assumed to be an average, healthy person, with indicative segment parameters, including muscle lengths, strengths, and masses. The motion data processed using this model is that of an affected H-OA population. Essentially, this study

evaluated altered gait patterns using a healthy walking gait model. H-OA participants have been reported to have smaller muscle size on the affected side than the contralateral side, and in some cases H-OA participants have smaller muscle size than control participants [52], which was not addressed in the model. This difference can affect estimations of JCFs when using a healthy model and unhealthy gait mechanics.

There is large variation in reported values of musculoskeletal modeling simulations, most likely arising from independent processing, different participants, participant walking speed, or data collection techniques. Different forms of collected data and measures provide different levels of accuracy within the modeling software. Musculoskeletal model choice decisions cause slight differences in the final estimations, which had been investigated in hip joint loading [31]. Processing workflow, such as using the CMC tool versus the Static Optimization tool, can cause variation in muscle force outcomes as well. Schellenberg et al. 2018 suggested that loading estimates from musculoskeletal models should be interpreted carefully as there is variation when it comes to reported results from modeling simulations [53]. Therefore, both direct and indirect validation are critical components to any modeling study.

Differences in JCF results in this study and previously published results could be the result of different data collection or analysis method. The *in vivo* measurements are direct measurements within the joint in contrast to the current study in which markers, motion capture, and force plates were used to develop a musculoskeletal model that estimated the JCF. Walking speed is another potential factor that could account for differences in JCF. Although all studies used self-selected walking speeds, it is up to the individual participants to determine what is comfortable for them. Faster walking speeds tend to correspond to higher loads on the joints and the average walking speed from this study (1.23 m/s) was faster than those of the publications used for comparison (0.83-1.17 m/s).

The model in this study is not accounting for differences in muscle parameters when comparing the average adult and an H-OA participant. H-OA participants have lower muscle force production in several muscle groups that could be accounted for when modeling, potentially lowering tracking errors in the pelvis and hip. They may have smaller muscle size in the affected limb versus the contralateral limb which could be addressed in the model as well.

This study's validity and acceptability could benefit from addressing participant-specific muscle parameters during model development.

In terms of the validation of this model, interpretation of results should be accepted with reservation. JCF output values have a much larger magnitude than those reported in previous studies and thus may not be numerically reliable. H-OA and control participants were analyzed the same way in this study; thus, they possess the same bias in JCF magnitude. For this reason, relative comparison of this model's results can be accepted.

Limb Differences in Joint Contact Forces in Hip Osteoarthritis Group

Abstract

People with osteoarthritis (OA) suffer from joint degeneration and pain that affects gait and mobility. Joint contact forces (JCF) are important for understanding loading within individual joints, especially in clinical populations. However, JCF cannot be measured directly without instrumented implants. Musculoskeletal modeling is a tool for researchers to estimate JCF without the need for invasive surgery. Therefore, the purpose of this study was to determine the effect of group (Hip OA (H-OA) n=10, control n=10) and side (affected, unaffected) on JCF during both the weight acceptance and push-off phases of level walking. It was determined that there was a significant difference in the knee and ankle JCF during the weight acceptance portion of stance and at all joints during the second half of stance when comparing the H-OA and control groups on the affected limb. A significant interaction between group and limb was found for the peak hip JCF timing (% stance) during the push-off portion of the stance phase ($p=0.009$). These results demonstrate that H-OA participants experience an earlier peak hip JCF during propulsion on their affected limb. Based on previous research in OA that has examined spatiotemporal measures, this finding suggests that H-OA participants may use step or stride length changes as a strategy to decrease or limit pain and loading on the affected limb. Knowledge of potential JCF differences in H-OA participants, such as timing of the peaks in either portion of the stance phase, could provide useful insight to clinicians and therapists to make decisions on how to proceed with treatment or rehabilitation programs.

Introduction

Osteoarthritis (OA) is the most common form of arthritis and a leading cause of global disability [1, 2]. At least 10% of adults over 60 and more than 20 million Americans are affected by OA [3]. The prevalence of symptoms increases with age, while symptom relieving treatment options decrease with age, contributing to OA's harmful impacts [3]. OA has many physical effects like joint degeneration, pain, swelling and stiffness, which affect gait and mobility [4, 12, 16, 35].

According to the World Health Organization, about 80% of people with OA have restricted movement and 25% are unable to perform daily tasks [8]. People with hip OA (H-OA) have greater gait asymmetries than healthy controls [4]. The greater asymmetry in these participants is likely due to gait compensations resulting from increased joint pain which can result in irregular wear and loading on contralateral and adjacent joints [9, 17].

Joint loading during normal activities have contributions from the ground reaction forces (GRF) as well as the muscles applying forces and moments to the joints [10]. The combination of these loads are known as joint contact forces (JCF). JCF are important because they provide and estimation of the forces that are applied on the articular surface, including stabilization of internal muscle forces, to better understand loading and strain within individual joints [54]. Musculoskeletal modeling is a potential tool for researchers to estimate and understand both external and internal forces acting on joints.

In H-OA, one specific model predicted a resultant peak force of 4.3 times body weight (BW) at the hip when walking at 1.36 m/s (4.9 km/h), consistent with the peak force of 4.8 BW walking at 1.39 m/s (5 km/h) measured by instrumented implants in another study [39, 45]. However, there is considerable variation in hip contact force measurements reported in the literature, possibly due in part to walking speed. A recorded mean peak hip contact force of 2.4 BW was reported with participants walking at about 1.08 m/s (3.9 km/h) [40]. Forces of about 5.5 BW at the hip were seen during jogging or fast walking, which occasionally spiked to about 7.2 BW [45].

In this study, a musculoskeletal model was used to estimate the JCF in the lower extremities of people with H-OA to understand the joint and limb specific impact that OA has on limb differences. JCFs were estimated for the hip, knee, and ankle of both limbs in a control population and a H-OA population. Hip, knee, and ankle peak JCF were calculated during the weight acceptance and propulsive portions of the stance phase, as well as the timing during the stance phase of the gait cycle of the peak hip JCF. The hip, knee, and ankle were all included when considering peak JCF because it has been reported that the presence of OA impacts other lower extremity joints [9, 18]. Both halves of the stance phase were examined based on findings in a 2015 study on the effect of hip, knee, and ankle OA, which determined that there were significant differences between control and OA groups in the ground reaction force (GRF) during the weight acceptance and propulsive phases of walking [18]. Only the hip was chosen

when considering the timing of the peak since reduced hip extension and flexion have been reported during walking in H-OA when compared to healthy controls [18].

Therefore, the purpose of this study was to determine the effect of group (H-OA, control) and side (affected, unaffected) on JCF during the stance phase of level walking. It was hypothesized that peak joint contact forces in the hip, knee, and ankle would be lower during the stance phase of walking in the H-OA population compared to the healthy control group.

Materials and Methods

In a previous study, motion capture and force plate data were collected from end-stage H-OA patients scheduled for a total hip replacement. A subgroup was identified for secondary analysis from these existing data. A power analysis was conducted using the peak GRF values to determine sample size. The results of this power analysis suggested that a sample size of 20 participants (10 control, 10 H-OA) were needed for 80% power. The participant demographic information is displayed in Table 2.1. All subjects had end-stage unilateral hip osteoarthritis as diagnosed by an orthopedic surgeon and were scheduled for a total hip replacement within one month of testing. Participants with contralateral hip pain, contralateral joint degeneration, previous total joint arthroplasty in the lower extremity, or history of neurological disorders were excluded from the study [49]. This participant sample was compared to a healthy control population, whose data were also previously collected [49].

Retroreflective markers were placed at 39 anatomical landmarks [49, 55]: sacrum (L5-S1), bilaterally on the acromioclavicular joint, lateral epicondyle, midpoint between the radial and ulnar styloids, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), greater trochanter, iliac crest, thigh, lateral knee, shank, lateral malleolus, posterior superior heel, second webspace (toes) in line vertically with the superior heel marker, posterior inferior heel, lateral heel, medial malleolus, medial femoral condyle, and first and fifth metatarsal head. The markers were recorded using an 8-camera motion capture system (Motion Analysis Inc, Santa Rosa, CA) sampling at 120 Hz. Subjects were asked to walk along a 40-m walkway. GRF data were collected using four AMTI force plates (AMTI, Inc, Watertown, MA) embedded in the floor sampling at 1200 Hz. Subjects were asked stand in anatomical position to record a static standing trial. Following the static trial, the medial malleolus, medial femoral condyle, iliac crest,

and first and fifth metatarsal markers were removed. Subjects were asked to perform 7 walking trials at a self-selected walking speed along a 40-m walkway. Trials were conducted barefoot to avoid changes in GRF due to footwear [55]. Height, weight, age, and Harris Hip score were recorded for each subject.

Following collection, data were processed using Visual3D (C-Motion, Bethesda, Maryland, USA) and OpenSim 4.0 [simtk.org, 23]. In Visual3D, an eight-segment model (feet, shanks, thighs, pelvis, and torso) was used to scale models to each participant based on proximal and distal marker positions on each body segment during a static, standing trial. Motion capture and embedded force plates provided data to produce inverse kinematic (IK) solutions from Visual3D using a global optimization algorithm from Lu & O'Connor [50]. Each participant must have 3 clean trials with at least 2 full steps on separate force plates, to be included in data analysis.

A 3D musculoskeletal model of the legs and torso was developed in OpenSim 4.0 [23] with 23 degrees of freedom and 92 Hill-type musculotendon actuators [29]. A Hill-type model can be represented by the force-length-velocity relation, and both passive and in-parallel contractive elements are assumed to contribute to the muscle force [47]. The OpenSim gait2392 model [29] was scaled using the Scale Model Tool, according to the participant demographics (height and mass) obtained during the data collection session. In OpenSim, the input kinematic and kinetic data were filtered at 6 Hz [44]. The IK solution and GRFs from Visual3D were input in the Reduced Residuals Algorithm (RRA) Tool to minimize experimental error and modeling assumptions and improve dynamic consistency [23, 51]. The model was adjusted based on the suggested mass adjustments and center of mass location of the torso from RRA results. The adjusted model and kinematic solution from RRA were then input in the Computed Muscle Control (CMC) Tool to estimate muscle forces and excitations that drive the model to track desired kinematics [23]. The CMC solutions were used to compute lower limb JCF using the Joint Reaction Analysis Tool. Data processing workflow in OpenSim follows the figure below (Fig. 2.1).

Figure 2.1: Data Processing Workflow in OpenSim

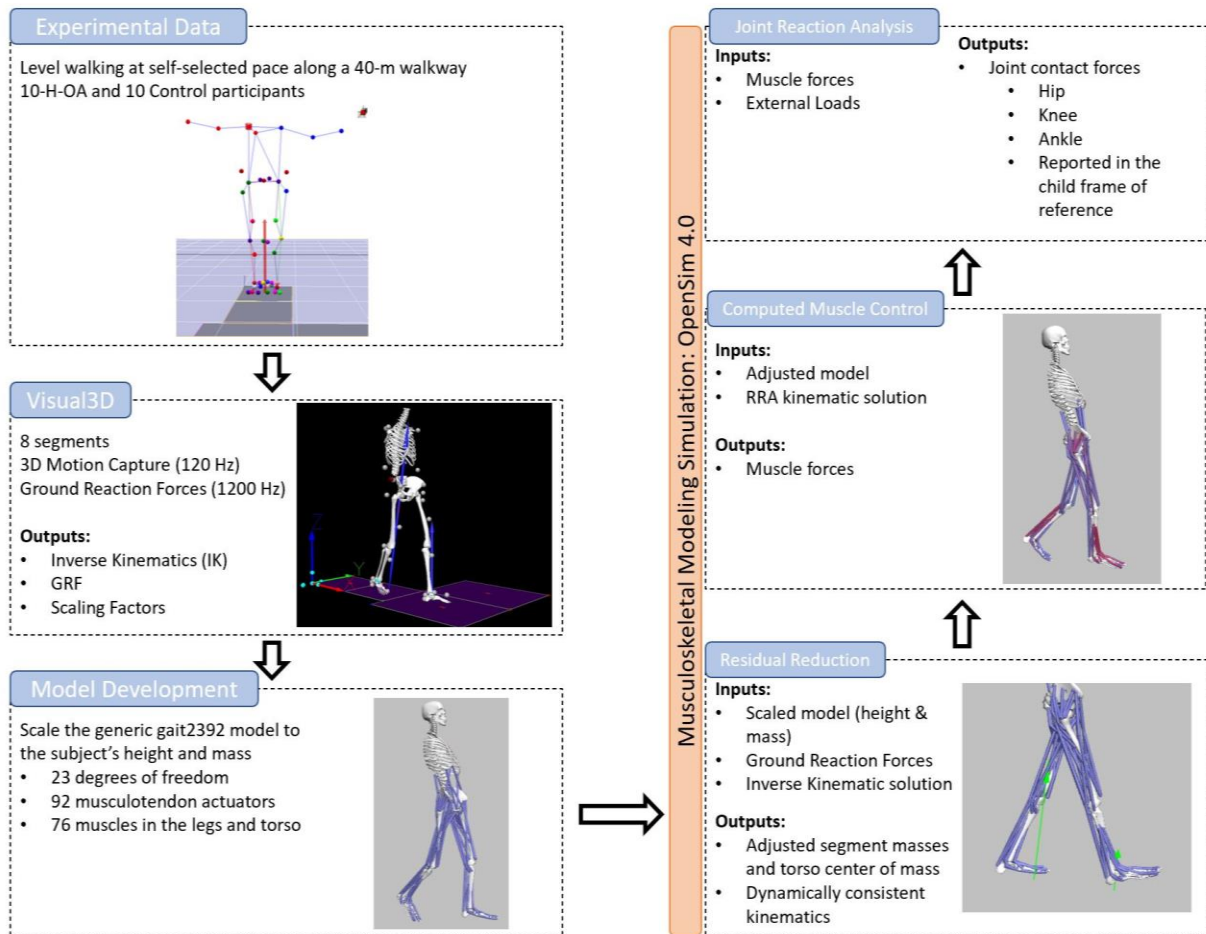


Figure 2.1: Workflow used for processing experimental data, model development, and generating walking simulations. In a previous study, level walking data were collected and processing in Visual3d. Motion capture and GRF data were sampled at 120 Hz and 1200Hz, respectively. IK, GRF, and scaling factors were exported from Visual3d for model development and processing in OpenSim 4.0 [23].

The stance phase was determined using MATLAB (MathWorks, Natick, MA) and was defined as the portion of the gait cycle between the subject’s initial contact with the force plate (heel strike) and the point at which the same foot was no longer in contact with the force plate (toe off). The peak JCF value was taken for each joint during the weight acceptance (first 50%) and propulsive (second 50%) portions of the stance phase and an average for each participant during each portion of the stance phase was determined.

All statistical analyses were performed using JMP (JMP Pro 16, SAS Institute Inc., Cary, NC). Independent sample t-tests were used to identify between-group (H-OA, control) differences in age, BMI, and walking speed (Table 2.1).

Table 2.1: Participant Demographics

	Control (n=10)	H-OA (n=10)	p-Value
Age	48.7 ± 7.75	63.9 ± 6.67	<0.001*
BMI (kg/m ²)	24.37 ± 4.41	27.18 ± 4.01	0.152
Walking Speed (m/s)	1.231 ± 0.175	1.266 ± 0.046	0.633

Table 2.1: Participant demographics. Both groups consisted of 10 subjects. The groups were statistically different in age. There was no statistical difference in body mass index (BMI) or walking speed. (indicates statistical significance (p<0.05))*

Distributions of JCF of each joint were assessed for normality using Shapiro-Wilk’s test (JMP Pro 16, SAS Institute Inc., Cary, NC). The peak ankle JCF residual distribution during the propulsive portion of the stance phase was identified as being non-normally distributed (p=0.034*), so nonparametric statistics were utilized for the ankle during the propulsive portion of the stance phase. Walking speed was not statistically different between groups, so walking speed was not included as a covariate. The peak JCF values were statistically evaluated using a 2x2 mixed model ANOVA for the hip, knee, and ankle for comparison of group and side-to-side limb differences between the H-OA and control groups during the weight acceptance and propulsive portions of the stance phase. Timing of the peak hip JCF, expressed as percent (%) of the stance phase, was compared between groups and limbs during weight acceptance and propulsion using a 2x2 mixed model ANOVA for each half of the stance phase. An alpha level of 0.05 was used to determine statistical significance. In the control group, the limbs were defined as dominant and nondominant with the right leg being defined as the dominant limb. In the H-OA group, the limbs were defined as affected and unaffected. For limb comparison between groups, the unaffected limb was defined as being comparable to the dominant limb.

Results

Joint contact forces for the hip, knee, and ankle were calculated using OpenSim 4.0 [simtk.org, 23] and stance phase was determined using MATLAB (MathWorks, Natick, MA) in both limbs for H-OA and healthy control participants (Fig. 2.2-2.4). See Appendix for figures including mean ± standard deviation.

Figure 2.2: Hip Joint Contact Forces for H-OA and Control Groups

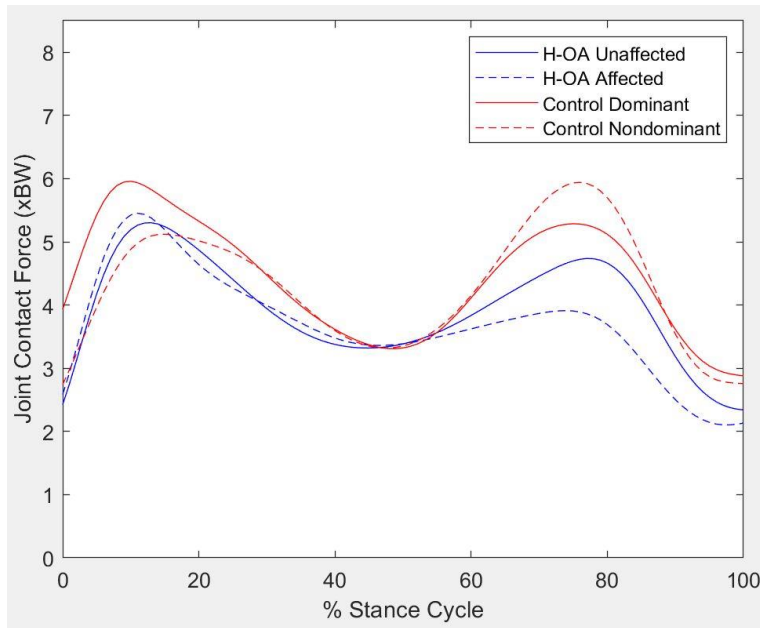


Figure 2.2: Joint Contact Forces at the hip for the H-OA group's unaffected and affected limbs, and the control group's dominant and nondominant limbs across stance phase.

Figure 2.3: Knee Joint Contact Forces for H-OA and Control Groups

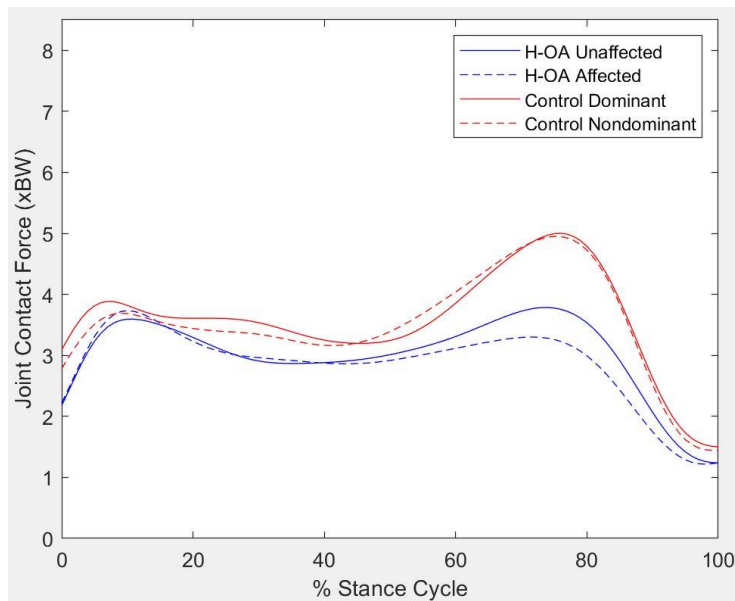


Figure 2.3: Joint Contact Forces at the knee for the H-OA group's unaffected and affected limbs, and the control group's dominant and nondominant limbs across stance phase.

Figure 2.4: Ankle Joint Contact Forces for H-OA and Control Groups

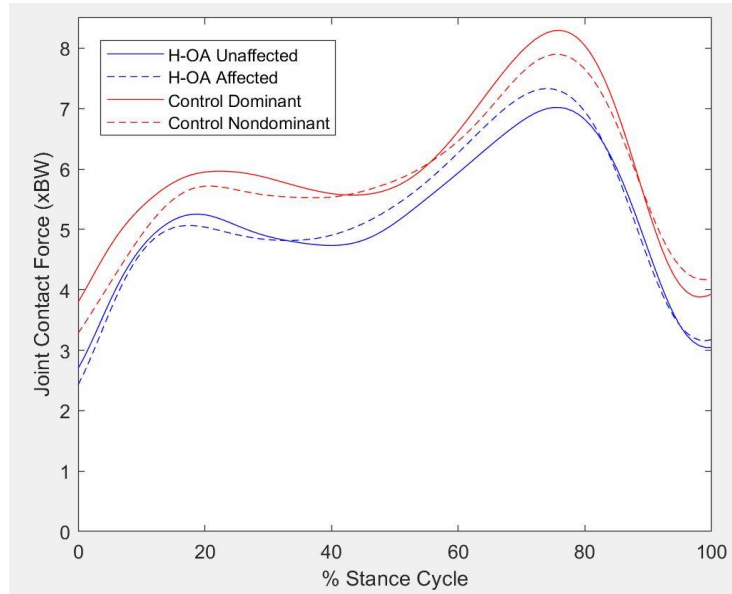


Figure 2.4: Joint Contact Forces at the ankle for the H-OA group’s unaffected and affected limbs, and the control group’s dominant and nondominant limbs across stance phase.

Group and limb differences were determined using a 2x2 mixed model ANOVA (Tables 2.2-2.3). No interactions were found at any joint during either half of stance phase. There was a significant difference between H-OA and control groups at the knee and ankle during weight acceptance, and at each joint during propulsion. No significant group difference was found at the hip during weight acceptance. There were no significant side-to-side limb differences at any joint during either half of the stance phase.

Table 2.2: 2x2 ANOVA results for the weight acceptance phase

	Group	Mean ± Std	p-value	Limb	Mean ± Std	p-value
Hip	Control	6.122 ± 0.19	0.141	Dominant	6.142 ± 0.19	0.107
	H-OA	5.717 ± 0.19		Nondominant	5.697 ± 0.19	
Knee	Control	4.409 ± 0.11	0.009*	Dominant	4.228 ± 0.11	0.706
	H-OA	3.989 ± 0.11		Nondominant	4.170 ± 0.11	
Ankle	Control	6.883 ± 0.24	0.008*	Dominant	6.430 ± 0.24	0.913
	H-OA	5.939 ± 0.24		Nondominant	6.393 ± 0.24	

Table 2.2: Results of the 2x2 mixed model ANOVA for the peak JCF values at the hip, knee, and ankle during the weight acceptance phase of stance. There was a statistical difference between groups at the knee and ankle. There were no side-to-side limb differences in any joints. (* indicates statistical significance (p<0.05))

Table 2.3 2x2 ANOVA results for the weight propulsive phase

	Group	Mean ± Std	p-value	Limb	Mean ± Std	p-value
Hip	Control	5.751 ± 0.11	<0.0001*	Dominant	5.148 ± 0.11	0.415
	H-OA	4.411 ± 0.11		Nondominant	5.014 ± 0.11	
Knee	Control	5.046 ± 0.11	<0.0001*	Dominant	4.486 ± 0.11	0.092
	H-OA	3.656 ± 0.11		Nondominant	4.216 ± 0.11	
Ranked	Control	24.6 ± 2.51	0.027*	Dominant	20.9 ± 2.51	0.823
Ankle	H-OA	16.4 ± 2.51		Nondominant	20.1 ± 2.51	

Table 2.3: Results of the 2x2 mixed model ANOVA for the peak JCF values at the hip, knee, and ankle during the weight propulsive phase of stance. There was a statistical difference between groups at each joint. There were no side-to-side differences in any joints. (* indicates statistical significance ($p < 0.05$))

During both weight acceptance and propulsion, there was no statistical difference in the timing (% stance) of the peak hip JCF between control and H-OA groups or between dominant and nondominant limbs (Tables 2.4-2.5). There was an interaction effect between groups and limbs during propulsion ($p = 0.009^*$), shown in Figure 2.5. The difference in timing between the unaffected and affected limb in the H-OA group was greater than the difference in timing between the dominant and nondominant limb in the control group.

Table 2.4: 2x2 ANOVA results for timing of peak hip JCF during weight acceptance

Group	Mean ± Std	p-value	Limb	Mean ± Std	p-value
Control	15.55 ± 0.82	0.097	Dominant	13.80 ± 0.82	0.192
H-OA	13.58 ± 0.82		Nondominant	15.34 ± 0.82	

Table 2.4: This table depicts the results of the 2x2 mixed model ANOVA for the timing of the peak hip JCF value during weight acceptance. There were no group or side-to-side limb differences. (* indicates statistical significance ($p < 0.05$))

Table 2.5: 2x2 ANOVA results for timing of peak hip JCF during weight propulsion

Group	Mean ± Std	p-value	Limb	Mean ± Std	p-value
Control	75.37 ± 0.76	0.157	Dominant	75.56 ± 0.76	0.077
H-OA	73.81 ± 0.76		Nondominant	73.61 ± 0.76	

Table 2.5: This table depicts the results of the 2x2 mixed model ANOVA for the timing of the peak hip JCF value during weight propulsion. There were no group or side-to-side limb differences. There was an interaction effect (Figure 2.5, $p=0.009^*$) (* indicates statistical significance ($p<0.05$))

Figure 2.5: Interaction effect of peak hip JCF timing during weight propulsion

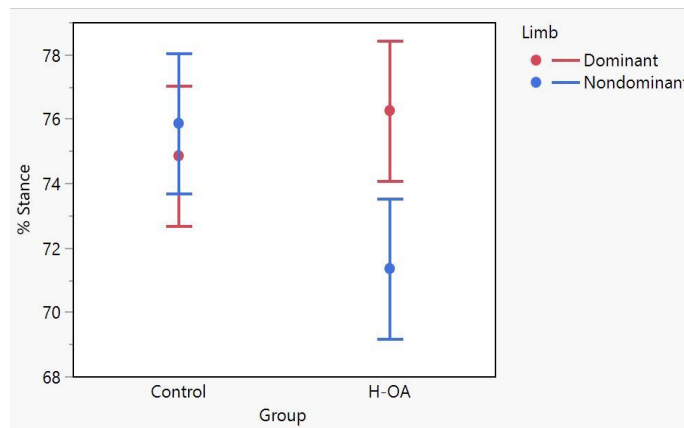


Figure 2.5: This figure shows an interaction between groups and limbs for the timing of the peak hip JCF during the second half of stance phase ($p=0.009^*$). The H-OA group had a larger side-to-side limb difference when compared to the control group.

Muscle force plots were produced for the major muscles spanning the hip, knee, and ankle that contribute to joint motion. Figures 2.6-2.8 provide plots of the muscle forces of individual muscles in muscle groups including the hamstrings, quadriceps, and hip flexors. More plots are available in the Appendix.

Figure 2.6: Muscle Forces in the Hamstrings

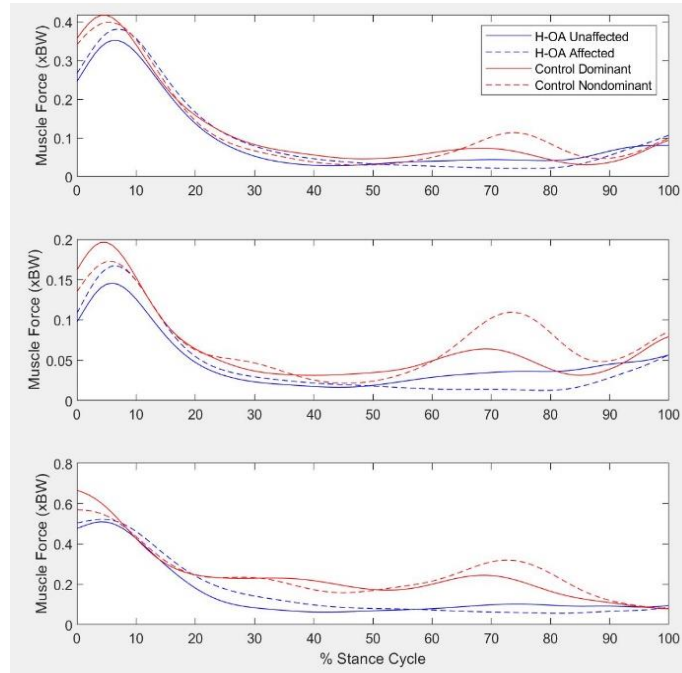


Figure 2.6: Muscle Forces from CMC solutions for the muscles that make up the hamstrings: biceps femoris long head, semitendinosus, semimembranosus, respectively. The H-OA group had lower muscle force production than the control group throughout stance phase.

Figure 2.7: Muscle Forces in the Quadriceps

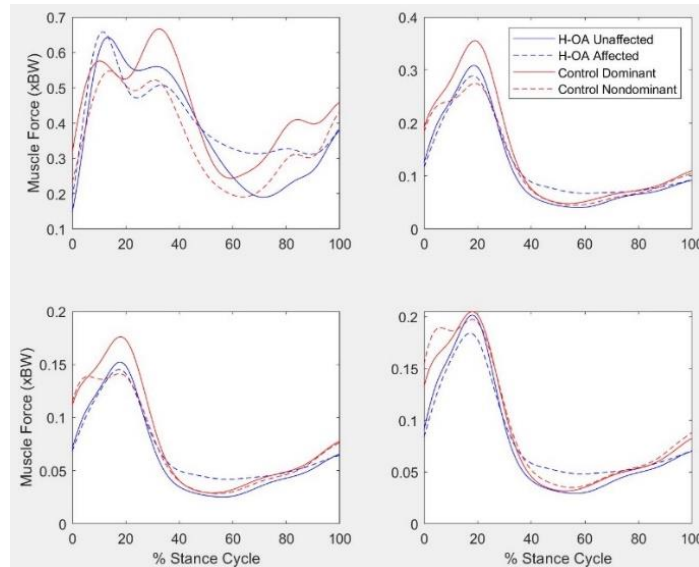


Figure 2.7: Muscle forces from CMC solutions for the muscles that make up the quadriceps: rectus femoris, vastus lateralis, vastus medius, vastus intermedius (from left to right, top to bottom). Throughout the majority of stance phase in the rectus femoris and all of stance phase for all others, the H-OA group had lower muscle force production than the control group.

Figure 2.8: Muscle Forces in the Hip Flexor Muscles

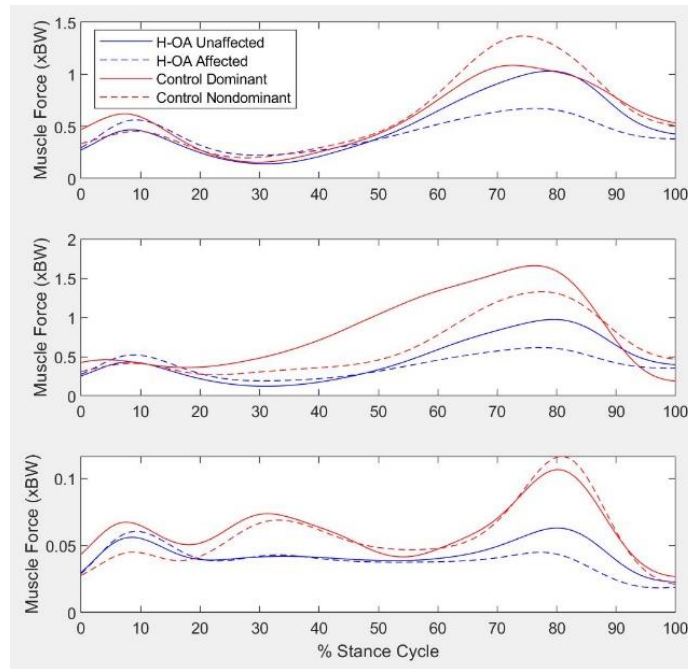


Figure 2.8: Muscle forces in the hip flexors muscles: psoas major, iliacus, sartorius, respectively. The H-OA group had a much lower muscle force production than the control group during the second half of stance phase in each of these muscles.

Discussion

The purpose of this study was to investigate group and side-to-side limb differences between H-OA and healthy control groups. The H-OA (n=10) and control (n=10) groups did not differ in body mass index (BMI) or walking speed (Table 2.1). OA generally affects adults over 60 years of age [3]; this was supported by the average age of H-OA participants recruited for this study. Additionally, the H-OA group was significantly older than the control group of healthy adults (control = 48.7 ± 7.75 , H-OA = 63.9 ± 6.67 , $p < 0.001$).

During weight acceptance, JCF in the knee and ankle were affected by group, while all joints were affected by group during propulsion. Conversely, limb was not significantly different for any joint during weight acceptance or propulsion. Although the timing of the first JCF peaks at the hip was not impacted by group or limb, interactions were found between group and limb during the propulsion phase of stance. By evaluating JCF in H-OA participants, these findings provide an improved understanding of joint loading as well as internal stability and muscle forces in H-OA participants.

As hypothesized, the H-OA group had lower peak JCF than the control group across all joints. This result agrees with previous reports in which H-OA groups exhibit lower hip and knee JCF when compared with healthy controls [33, 37, 56]. Lower JCF could be attributed to lower muscle force production and muscle strength, as muscle forces are the major contributors to JCF [39]. H-OA participants in this study saw slightly lower muscle force production through stance phase in a large majority of the muscles spanning the hip, knee, and ankle (see Appendix). Muscle weakness on the affected limb versus contralateral limb and in H-OA participants versus control participants was reported in the hip adductors and abductors, and hip and knee flexors and extensors [52]. These muscle groups were compared between limb and group. Hip flexors, extensors, adductors, quadriceps, and hamstrings show lower muscle forces in the H-OA group compared to the control group, and in the affected limb compared to the unaffected limb. The largest differences between H-OA and control groups were right around where the JCF peaks occurred, particularly the hamstrings, quadriceps, and hip flexors (Fig. 2.6-2.8). In a review article on muscle weakness in H-OA, it was found that H-OA participants have lower muscle strength in the affected limb relative to healthy controls [52]. The lower peak JCF for H-OA participants in this study can also be contributed to OA's known influence on joint degeneration, gait, and mobility changes, and inhibiting muscle strength [4, 35, 52]. This is consistent with the fact that lower force and power production from the muscles results in lower overall joint contact forces.

Despite expectations, no side-to-side limb differences were found in any of the analyses performed. This is an interesting finding because OA populations have been found to exhibit asymmetry in a variety of gait parameters [4, 9, 18, 37, 57]. This result could be attributed to sample size since it may be possible that side-to-side limb differences could be seen in the H-OA population with a larger sample size. There are many limitations with musculoskeletal modeling approaches, considering specific muscle parameters such as strength, length, and activation are unknown in this study. As a result, the model is left to make assumptions about these parameters. Muscle parameters could be provided through EMG collection or medical imaging such as MRI scans, depending on what researchers have available. Current musculoskeletal models base muscle model parameters primarily on young male cadavers [46], however this lacks individuality in participant fitness and strength. Knowledge of muscle parameters on an individual level would provide more accuracy in variables such as muscle activation behavior or

muscle fiber lengths of the participants. Musculoskeletal models could produce more accurate muscle force calculations, and thus JCF estimations, if these variables were better parameterized on a participant-specific basis.

Although not statistically significant, the nondominant/affected limb had lower peak JCF than the dominant/unaffected limb in both groups. In the H-OA group, this is likely a compensatory pattern meant to decrease or limit loading on the affected limb to reduce the joint pain associated with walking. There is evidence of lower muscle strength in the affected leg compared to the unaffected leg in a H-OA population [52], which could explain the lower peak JCF values seen in this study.

There was also an interaction between group and limb when considering the timing of the peak hip JCF during weight propulsion. The H-OA group had a greater difference between unaffected and affected limbs than the difference between the dominant and nondominant limbs in the control group. In the H-OA group, the affected hip reaches its peak JCF earlier in the stance phase than the unaffected hip. This could be due to the need for H-OA participants to compensate for the affected limb while walking. Earlier JCF peaks can be interpreted as a strategy to reduce pain in the affected limb; this can be observed as participants try to decrease loading on a joint. Reaching the peak JCF in the second half of stance is likely an attempt to push the affected limb through the propulsive phase quicker and possibly decrease contact time on the affected limb. By manipulating the amount of time the affected limb is in contact with the ground, H-OA participants can mitigate joint pain associated with the external forces contributing to JCF. This assumption follows previous literature, which found increased step and stride lengths [18], decreased swing time and stance time on the affected limb [57], and lower peak vertical GRFs [58, 59] in an ankle OA population.

JCF are typically normalized by participant body weight [27, 32, 40, 45], however, in some cases, H-OA participants may be overweight or obese, they may carry their weight distribution differently [13]. Overall BW normalization may not have been the best decision for this participant population, as normalization essentially scales the participants to a standardized body size and may not reflect the size of the joint of interest [62, 63]. This population may have been more effectively and reliably represented if this study normalized measures by mass distribution, overall peak JCF value, baseline leg strength determined by an isolating exercise (e.g. leg press), or size of joint.

The gait2392 model has some potential limitation, including that the model has no patella, a pure hinge joint at the knee, and lacks ligament representation [23]. The pure one plane hinge joint allows the knee to flex and extend, with no abduction or adduction motion, however the knee realistically would have some minimal motion in this plane as well as axial rotation that is not being expressed in the model. JCFs are a combination of external and internal forces and moments on the joints, including cartilage and ligament contributions. Ligaments are important to consider in the biomechanics, as they are the connection between bones within joints. Ligaments can adapt to changes in mechanical behavior and loading by adjusting their size or material property [61]. Diseases, like OA, and age both impact the way participants walk and functionality of the joints, so ligament contribution would likely differ from an average healthy participant. The lack of ligament representation in the model may affect the JCF calculations, as the model ignores this aspect of contribution. This limitation should affect both H-OA and control groups.

As previously mentioned, a potential limiting factor in the results of this study could be the sample size. A power analysis was performed to determine sample size of 20 subjects (10 control, 10 H-OA) based on GRF data; however, a larger sample size may be necessary to find side-to-side limb differences based on the results of the modelling process. Perhaps another limitation of this study is that the data used for computation was previously collected with a different purpose. This data was not collected in a way that was intended for use in OpenSim; future studies may benefit from understanding how data can be collected to be compatible with OpenSim. Future studies should include collection of muscle parameters – length, strength, activation, co-contraction. EMG-assisted neural control solutions can be more consistent with experimental data than simply musculoskeletal modeling solutions [32].

These findings will benefit H-OA participants and treatments available to them by providing further understanding of how this population compensates in terms of JCF of major load bearing joints during level walking. This information could provide useful insight to clinicians and therapists to make more informed decisions on how to proceed with treatment or rehabilitation programs, potentially delaying the need for invasive joint replacement surgery.

Conclusions

Musculoskeletal modeling was employed to estimate joint contact forces (JCF) in hip osteoarthritis (H-OA) participants and determine group and/or side-to-side limb differences when compared with healthy control participants. JCFs provide insight regarding joint loading from ground reaction forces (GRF), muscle forces, and joint moments [10], but there is an inability to measure JCF directly without instrumented implantations. There is large variation in reported JCFs from musculoskeletal modeling simulations. The reported JCFs typically range anywhere from 1.5 to 4 times the participant's body weight (BW) during self-selected speed level walking (walking speed ranging from about 0.87 to 1.4 m/s) [27, 31, 32, 35], therefore model validation and simulation quality analysis was also completed.

The first purpose of this research was to develop and validate a method utilizing musculoskeletal modeling in which lower extremity JCFs were calculated at the hip, knee, and ankle in H-OA and healthy control samples using OpenSim 4.0 [simtk.org, 23]. Metrics for validation included comparison of JCF trends and peak values to published modeling calculations of JCFs and *in vivo* JCF measurements from instrumented implants in a similar participant sample. Simulation quality was also assessed by analysis of residual forces and moments, tracking errors between inverse kinematics (IK) and computed muscle control (CMC) solutions, and reserve actuators.

Limitations arise when comparing experimental data for preoperative participants with *in vivo* measurements from post-operative implant participants. Preoperative participants are likely experiencing pain that post-operative participants may not be experiencing. People that have undergone a surgical procedure may have some differences in walking kinematics, joint function, muscle behavior, and walking speed, which cannot be accounted for during model validation.

The second purpose of this study was to investigate group and side-to-side limb differences between H-OA and healthy control participants through the analysis of JCFs during the stance phase of level walking. Although there were no significant side-to-side limb differences in peak JCFs at any joints, there was a significant interaction between group and limb in the timing (% stance) of the peak hip JCF during the push-off phase of gait. The H-OA group had greater differences between unaffected and affected limbs when compared to the dominant to non-dominant limbs in the control group. Earlier JCF peaks in the affected limb can be interpreted as a strategy to reduce pain by manipulating contact time as well as step length, stride length, and

exhibiting a lower vertical peak GRFs in the affected limb, which follows previous literature on participants with ankle OA [18, 57, 58, 59].

Peak JCF group differences existed in the knee and ankle during weight acceptance, and the hip, knee, and ankle during weight push-off. H-OA participants had lower peak JCF which could be attributed to lower muscle force production in many of the major lower extremity muscle groups – hip flexors, extensors, adductors, quadriceps, and hamstrings. Muscle forces are a major contributor to JCF [39] and these findings align with previous findings of muscle weakness in H-OA participants [52].

A limitation to this research could be that potential muscle weakness and smaller muscle size in H-OA participants was not accounted for in model development. Both H-OA and control participant analysis started with the same generic gait2392 model [29], scaled to participant height and mass and not muscle parameters. The gait2392 model is assumed to be an average, healthy person and the motion data processing using this model is that of an affected H-OA population. This may have affected the JCF calculations and would be advised to consider for future studies.

Although not statistically significant, H-OA participants experienced lower JCFs in the affected limb compared to the contralateral limb. Understanding that H-OA participants are experiencing lower JCFs and lower muscle force production spanning major load bearing joints in the affected limb enforces the need for rehabilitation and treatment programs to target strengthening the hip, quadriceps, and hamstrings, potentially delaying the need for surgery. The knowledge of group and limb differences could allow for the development of new and/or improved treatment options for patients suffering from OA. Therapists can adjust rehabilitation programs to better fit their OA patients, while clinicians can be provided with information that will allow for more informed decisions on alternative treatments before suggesting surgery.

Overall, interpretation of results should be accepted with reservation. JCF output values have a much larger magnitude than those reported in previous studies (0.5 to 3.0x BW larger) and thus may not be numerically reliable. The general trend of calculated JCFs throughout stance phase was similar to that of previously published data. There is some level of confidence in the walking simulation outputs as residuals and tracking errors remain low for the entirety of stance phase in most cases. Vertical force residuals were outside the accepted range of < 0.02 . Given that the overall reserve actuators were low, the muscles in the model were able to reproduce the

experimentally observed walking pattern. H-OA and control participants were analyzed the same way in this study; thus, they possess the same bias in JCF magnitude. For this reason, relative comparisons of this model's results can be accepted.

The biggest limitation with this research was the fact that the data used was previously collected for a different purpose, not intended to be used in OpenSim. In order to address this limitation, future researchers should collect new data intended with the requirements of OpenSim in mind. When assigning the study, this should include 3D motion capture, ground reaction forces, and EMG data – either for input into outcome metric calculations in OpenSim or for use to compare muscle calculations as validation. Researchers would also benefit from understanding where differences may lie between their participant samples. For example, H-OA participants have muscle parameters differences – smaller muscle size, muscle weakness – due to the effects of OA [52], but also age-related muscle changes, as this H-OA sample is significantly older than the healthy control participants. As people age, muscle changes – such as decrease in muscle mass and muscle protein synthesis, and denervation of muscle fibers – can affect activity level and quality of life [64].

In an ideal research environment, future studies should adjust muscle parameters to match those of the participant sample under observation. H-OA participants tend to have smaller muscle size and lower muscle strength [52]. Muscle size can be collected using MRI or CT scans and muscle excitation and activation can be collected with EMG. SimTK suggests collecting EMG data for as many muscles as possible as a best practice when preparing experimental data collection for simulations using OpenSim [29]. Additionally, a larger sample size may be necessary to find side-to-side limb differences. A power analysis based on GRF data determined a sample size of 20 subjects was sufficient for 80% power, however JCF results may have yielded lower power, thus a larger sample size is recommended. The combination of these suggested changes should appropriately advance the accuracy and validation of this research.

Appendix A

Joint Contact Forces

Figure A.1: Hip Joint Contact Force Variability for H-OA and Control Participants

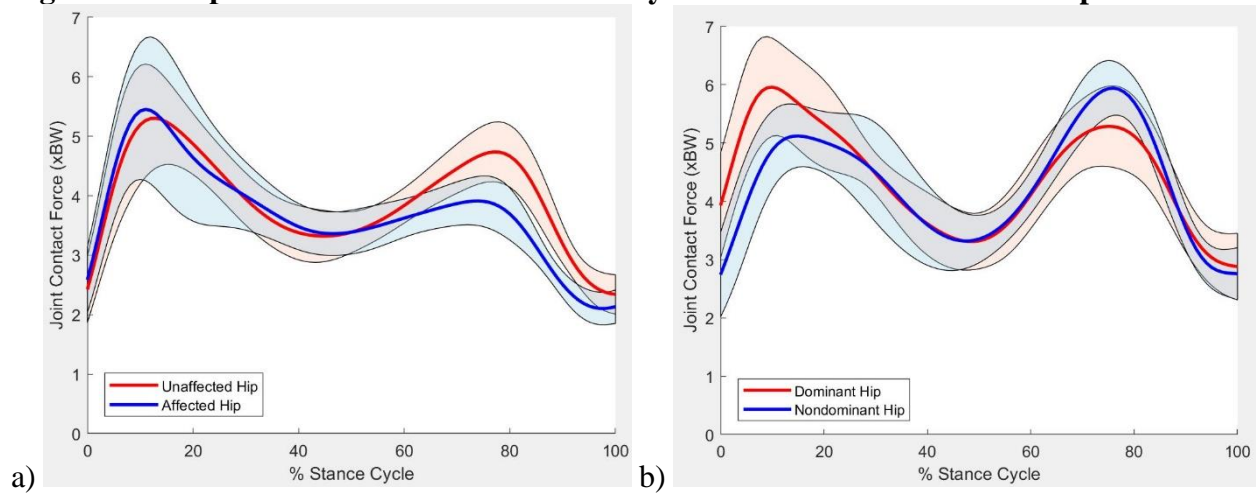


Figure A.1: Hip Joint Contact Force mean \pm standard deviation; a) H-OA participants; b) Control participants

Figure A.2: Knee Joint Contact Force Variability for H-OA and Control Participants

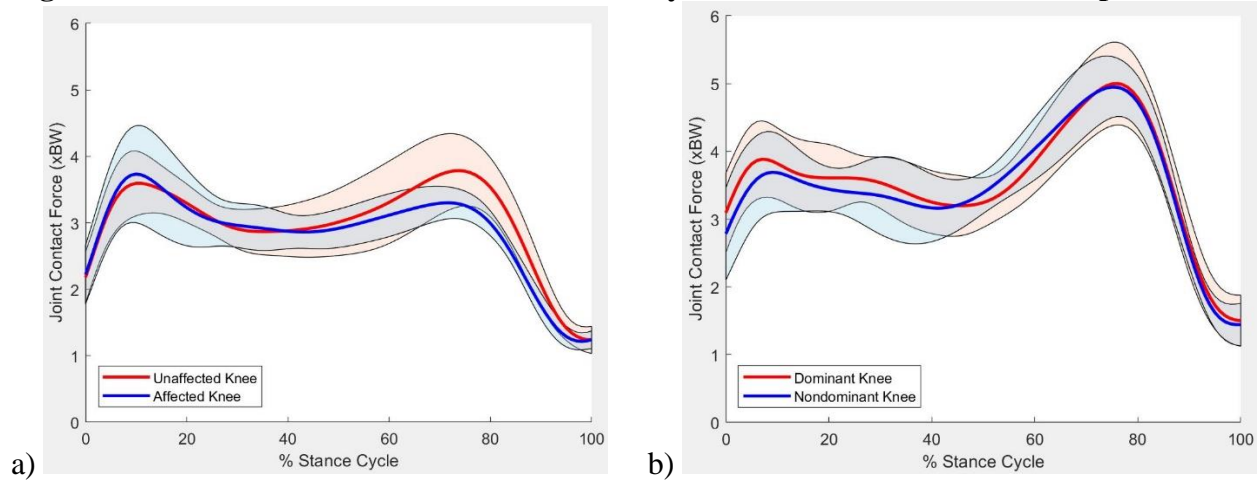
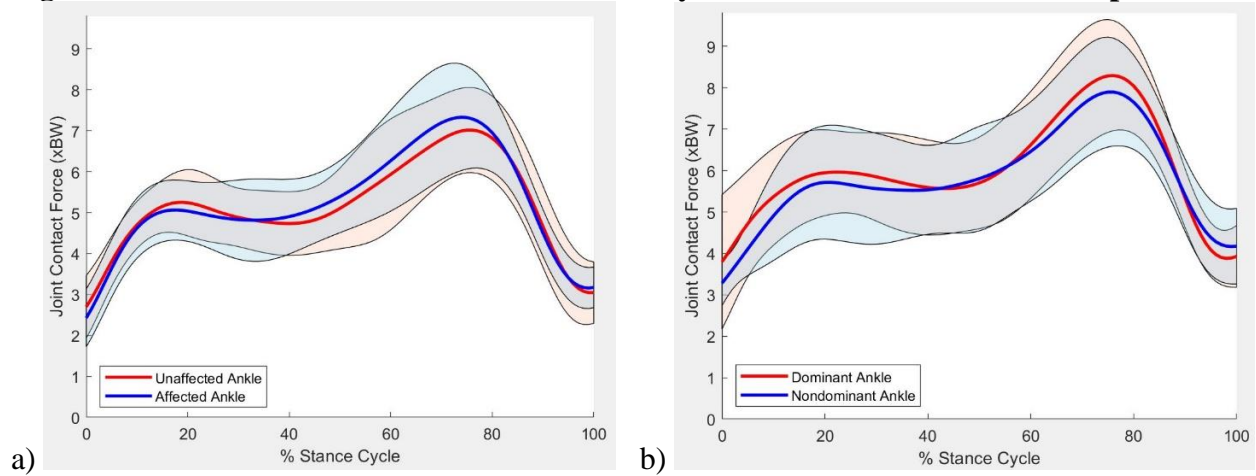


Figure A.1: Knee Joint Contact Force mean \pm standard deviation; a) H-OA participants; b) Control participants

Figure A.3: Ankle Joint Contact Force Variability for H-OA and Control Participants



Muscle Forces

Figure A.4: Gluteal Muscle Forces

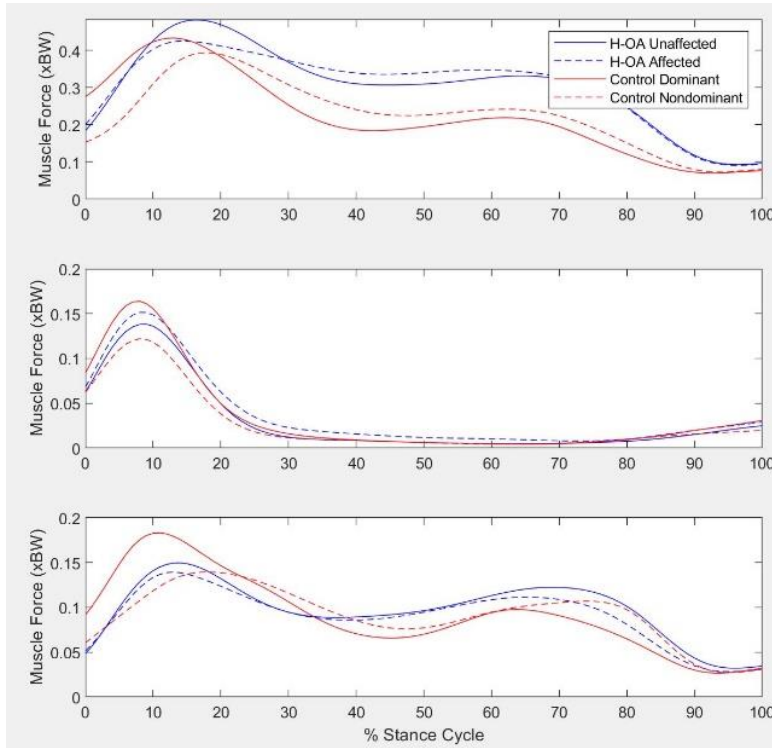


Figure A.4: Gluteal muscle forces: gluteus medius, gluteus maximus, gluteus minimus

Figure A.5: Hip Adductor & Hip Flexor Muscle Forces

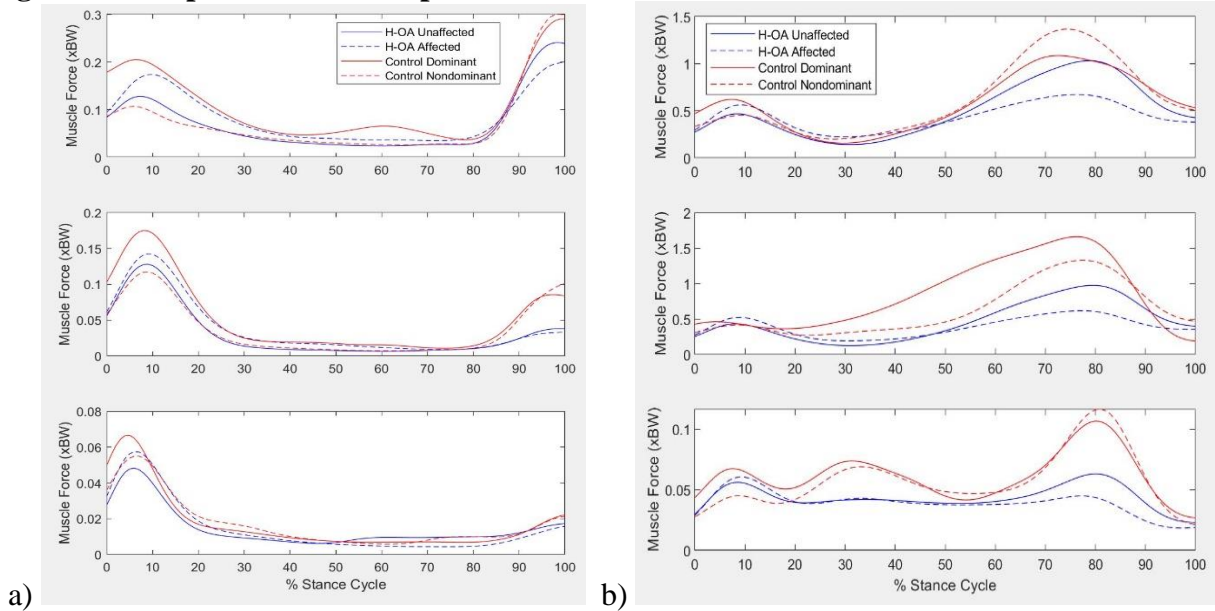


Figure A.3: Force in the muscles spanning the hip. a) hip adductor muscles (adductor longus, adductor magnus, gracilis); b) hip flexor muscles (psoas major, iliacus, sartorius, respectively). The H-OA group had a much lower muscle force production than the control group during the weight acceptance phase in the hip adductors & during the push-off phase in the hip flexor muscles

Figure A.6: Muscle Forces in the Hamstrings

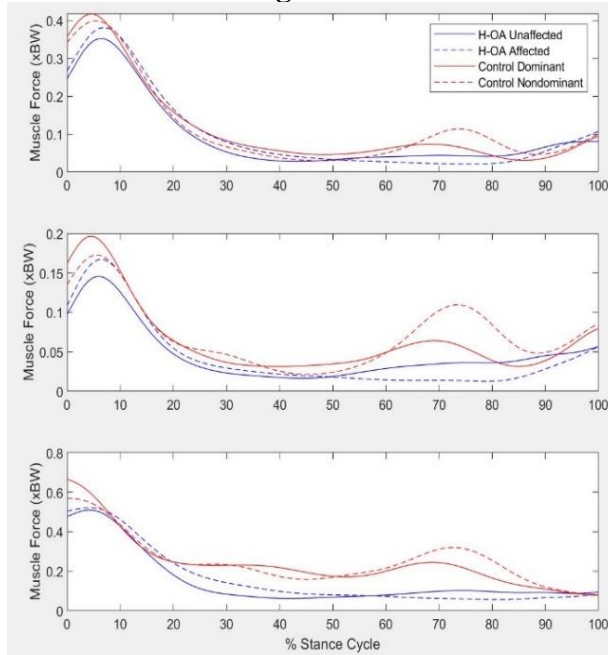


Figure A.6: Muscle Forces from CMC solutions for the biceps femoris long head, semitendinosus, semimembranosus, respectively. The H-OA group had lower muscle forces than the control group throughout stance phase. The affected limb also had lower muscle force production than the unaffected limb.

Figure A.7: Quadriceps Femoris Forces

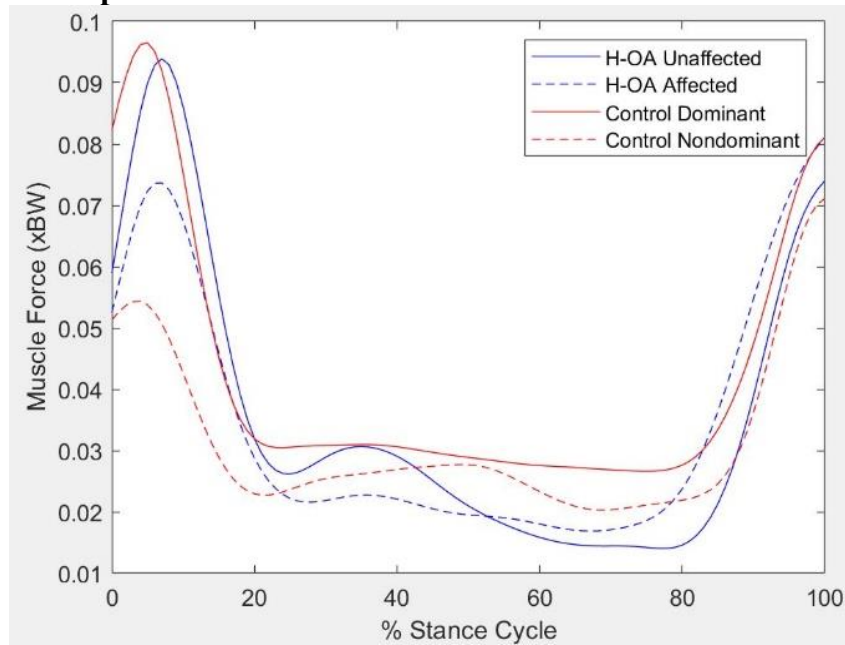


Figure A.7: Muscle forces in the quadriceps femoris from CMC solutions.

Figure A.8: Forces in Quadricep Muscles

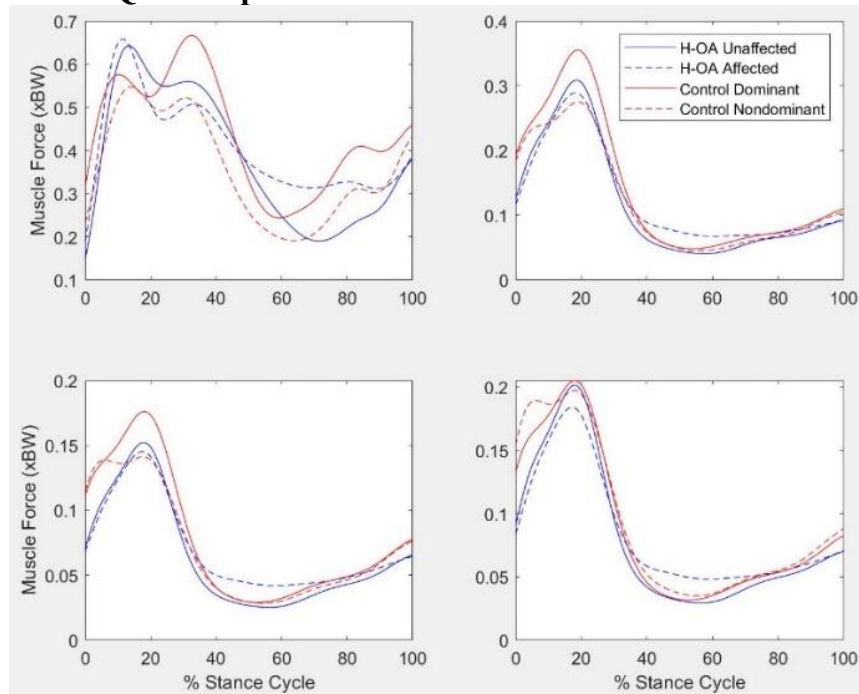


Figure A.8: Muscle forces from CMC solutions for the muscles that make up the quadriceps: rectus femoris, vastus lateralis, vastus medius, vastus intermedius (from left to right, top to bottom). Throughout the majority of stance phase in the rectus femoris and all of stance phase for all others, the H-OA group had lower muscle force production than the control group.

Figure A.9: Forces in Ankle Muscles

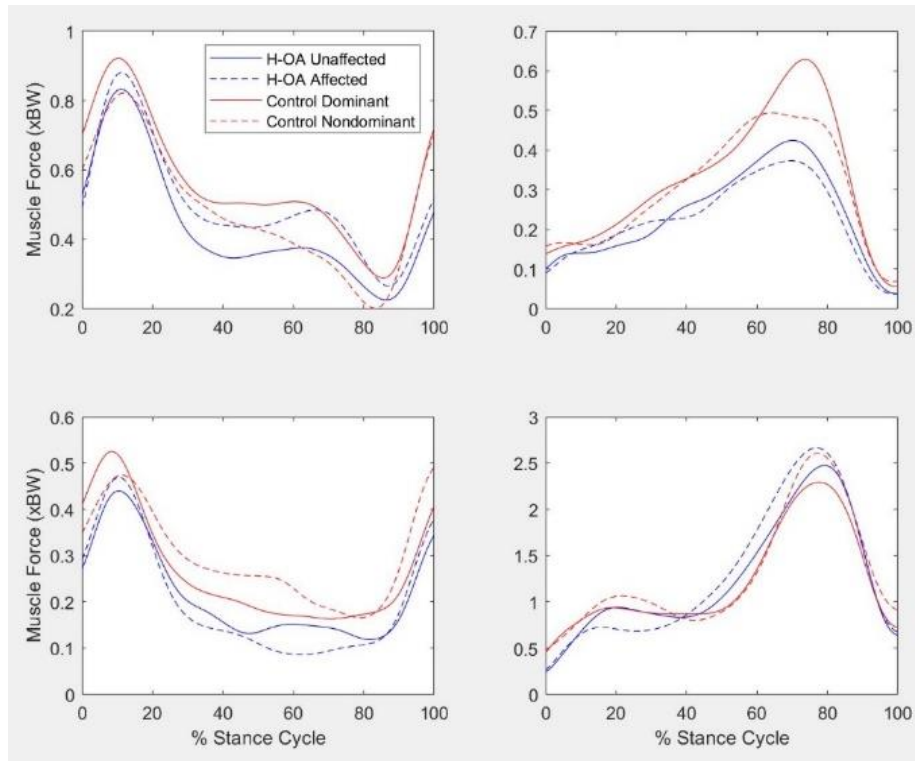


Figure A.9d: Muscle forces in the muscles that span the ankle: tibialis anterior, lateral gastrocnemius, extensor digitorum longus, soleus (from left to right, top to bottom)

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