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Development of a Computational Framework for Estimating Knee Joint Contact Forces in Running

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Abstract

The prevalence of running as a form of exercise and the necessity of walking for simple locomotion obscure massive forces and moments within the body. An especial area of concern is the knee, as common among these injuries is knee pain as a result from high impact on the ground or ground reaction forces. These forces are altered by the foot strike pattern of the individual; in this study, either rearfoot strike (RFS) or forefoot strike (FFS). This alteration will impact internal forces conducted upward through the body. Given the complexity of the motion of running and the forces involved, is it useful to apply computational modeling to study the underlying mechanical aspects of walking and running. The OpenSim modeling software provides many of the resources required to create and test such actions. A generic musculoskeletal model is scaled to patient specifications, and using marker data and ground reaction force data, kinematics of the captured movement and individual muscle forces are calculated. A simulated model and computational tools allow for measurements physically impossible *in vivo*; specifically, the compressive tibiofemoral force that may have adverse effects on tissues such as articular cartilage or menisci. While the peak compressive force and ground reaction force (GRF) was found to be higher for FFS than RFS, the impact transient at initial contact was significantly reduced in FFS in both GRF and knee joint contact forces. These findings further explore the impact of gait alterations on internal loading and may provide evidence for future studies examining related parameters such as muscle activation or joint kinematics in association with joint force.

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Introduction

Running is one of the most popular forms of exercise. In a survey encompassing nearly all common forms of physical activity, it was found that 8.8% of Americans participate in running [1]. The frequency and intensity of the exercise is such that between 37-56% of runners experience injuries annually [3]. Part of this is the nature of the compressive loads conducted upward through the knees. Walking creates ground reaction forces (GRFs) of 1.2 times bodyweight and exposes the knee joint to a large compressive force of up to 3 times bodyweight [4-6]. Running produces ground reaction forces (GRFs) up to 2.5 times bodyweight for most people, and a corresponding compressive tibiofemoral force of 8-15 times bodyweight [7, 8]. This compressive force can be understood as the product of several anatomical and environmental factors. As there are so many factors that contribute to the action of running, there are several parameters that may be adjusted to decrease joint force and thereby improve running technique. To investigate the effect that a particular alteration (in this case, foot strike pattern) will have on this internal loading, an appropriate model that accurately reflects the state of a subject is crucial.

Measuring knee loading is a nontrivial process. In patients who have had a knee replacement, technology exists to measure tibial force in vivo by way of a force transducer within the prosthesis [9, 10]. This is clearly less preferable for healthy populations, where such devices may be uncomfortable or invite unnatural movement due to the invasive nature of the instruments involved. To obviate the need for direct measurement, surrogate measures of medial tibiofemoral compartment loading have been developed, such as the knee adduction moment. Greater knee adduction moments can lead to greater medial contact force loading on the knee, which are linked to knee disorders such as medial tibiofemoral osteoarthritis [11]. The accuracy can be increased by including information from concurrent indicators such as knee flexion moment [12]. Still, knee abduction measurements are not a direct indicator of medial compartment loading and have had mixed success [11, 13]. Computational modeling addresses these constraints and inaccuracies by avoiding the need for invasive measurement devices or surrogate indicators of contact force.

Previous studies have investigated the differences between foot strike patterns in the biomechanics of runners [14]. There are various types of foot strike patterns, and the difference can be easily discerned by viewing the strike in the sagittal plane (Fig. 1). The common type is rearfoot strike (RFS), used by the vast majority of long-distance shod runners [15]. In RFS, the heel makes initial contact with the ground, and the foot rolls along the ground, culminating with a “toe-off” at the end of the strike. RFS implies strong distal activity of dorsiflexor muscles.

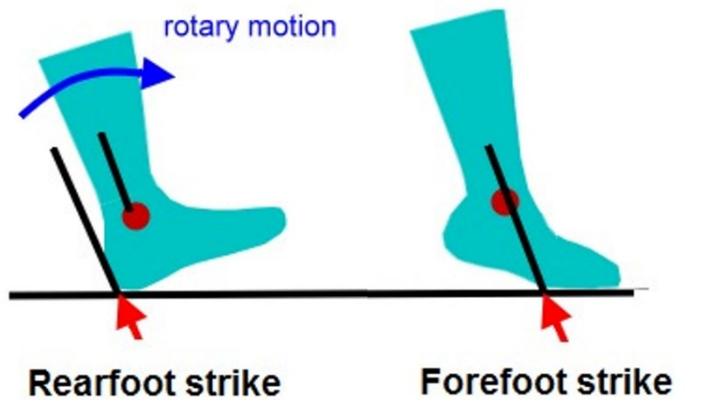

Fig 1: Difference in foot strike patterns.
From https://fl.milesplit.com/articles/112223/heels-or-toes-what-is-the-best-way-to-run [2].

The opposite is true for a forefoot strike (FFS), where the toe area is the first part of the foot to contact the ground, and results in greater plantarflexor muscle activity [16]. While FFS is not an instinctive gait for most shod runners, there is evidence to support the idea that barefoot runners tend toward FFS [17]. Benefits have been linked to using an FFS over a RFS pattern: Milner et. al advocated FFS in order to prevent injury, as the strike pattern was linked to lower vertical ground reaction forces [18], which are used as an approximation for lower extremity loading [14]. More recent and advanced models have also approached the problem of foot strike pattern alteration on patellofemoral joint pain, finding a peak reduction of 27% when subjects changed from RFS to FFS [19]. The ground reaction force profile is also observed to differ between RFS and FFS. Specifically, the impact transient that occurs when a runner makes ground contact with a RFS is practically eliminated when switching to FFS [17]. The knee performs less work in FFS than in RFS, but this work seems to be transferred to a higher-performing ankle [20]. It has also been proposed that FFS exercises both greater elastic energy storage and return in significant nearby structures such as the Achilles tendon and medial longitudinal arch. The force concentration difference has broader effects on the gait cycle, as FFS users displayed reduced stride length and gait speed [16]. FFS kinematic differences are not significant between learned and intrinsic FFS users [20], supporting the idea that FFS is an intuitive strike pattern. For these reasons, FSP modification is a promising area for injury prevention and treatment.

The purpose of this study is to compare compressive tibiofemoral force during RFS and FFS running. A multibody simulation of lower extremity kinematics and kinetics will be used to estimate joint coordinates and muscle activations as parameters. Using these values, the maximum overall force as well the impact transient will be compared.

Methods

Participants

The data for this study were collected from twenty-four healthy females between the ages of 18-35 with a habitual RFS. Participants with a lower extremity injury within the past year were excluded. All participants gave informed consent as approved by the university's Institutional Review Board [21].

Protocol

Wearing New Balance 10v1 Minimus Trail-Running shoes, participants had markers placed in significant areas: pelvis and right leg (thigh, shank, and foot). The first trial was a static trial for scaling purposes, followed by five trials of RFS running. Participants were then given a simple command, "contact the ground with the ball of the foot", to convert to an FFS pattern for which five trials were collected. Marker trajectories were tracked by an 8-camera Vicon Motion Capture System (Oxford Metrics, UK) at 150 Hz. Ground reaction forces were captured at 1500 Hz by an inground force plate (Bertec Corp., Columbus OH, USA). The marker and force data were collected during right leg stance phase. In MATLAB (Mathworks, Natick, MA, USA), marker data were low-pass filtered by a 4^{th} order Butterworth filter at 10 Hz. Force plate data were filtered in the same way at 50 Hz.

Data Analysis

The data were analyzed using OpenSim software. OpenSim is an open-source software for building biologically accurate models to simulate musculoskeletal dynamics. An OpenSim model is a collection of components that represent a biological system; in this case, the model consists of skeletal rigid bodies and muscles that reflect force-length-velocity relations of real muscle [22]. The behaviors and interactions of these components produce a system subject to the individual constraints of the components [23]. This system occupies a number of states—consisting of muscle activation, joint angles, and other model properties—that are collected across a time interval by OpenSim's computational tools. The experimental data inputs and processing steps can be seen in Fig. 2.

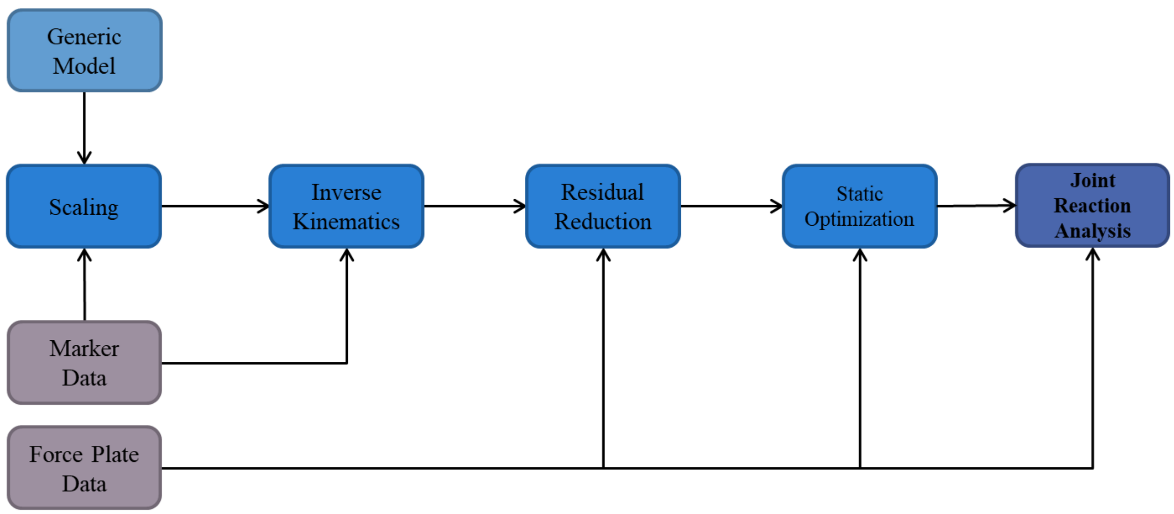

Fig. 2. Data processing pipeline.

The model used for this study was based on a model used by Hamner et. al to investigate muscle activity in running [4]. The representative markers from the data collection were added to the model and the structures for which no data were collected (upper extremity and left leg) were removed. The resulting model has13 degrees-of-freedom with 43 muscle-representing actuators, seen in Fig. 3 This is the generic form of the model to which personalizing attributes will be added to reflect individual subjects.

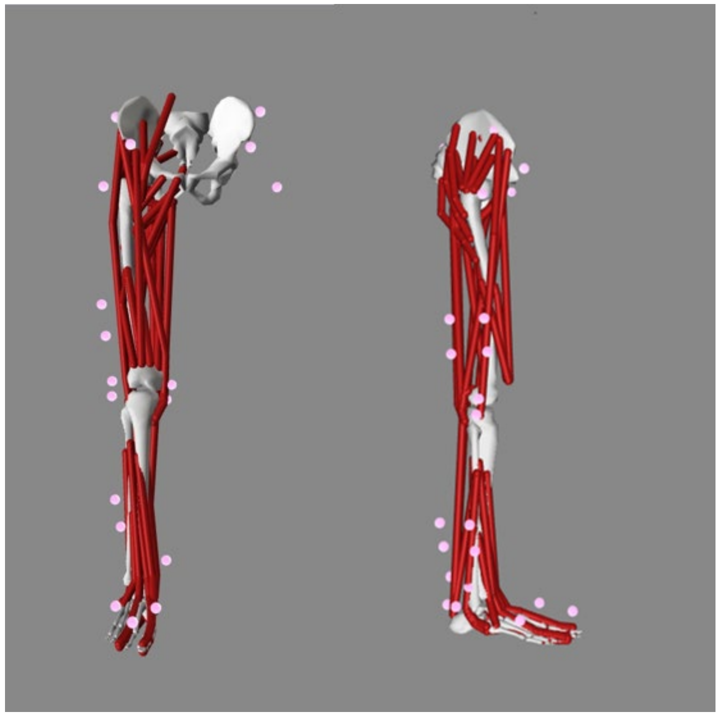

Fig. 3. Frontal and sagittal plane views of the generic OpenSim model

The first step to generate an accurate model is scaling. The same markers are present virtually on the OpenSim model as were present on the human participant at the time of data collection. As opposed to using manual scale factors, the Scale tool uses pairs of markers to determine individual scale factors [23, 24]. These scale factors are determined from an average across a static trial collected while the participant remains still. The Scale tool uses these scale factors to redefine body geometry. The same scale factor is used to scale mass centers as well as the mass and inertia tensors that define each segment. Muscle attachment points and properties such as optimal fiber length are scaled next. The muscles themselves also undergo scaling, although iteratively this task is completed after components that do not depend on length. After the parts of the model are scaled appropriately, the virtual markers on the model before the tool is run are moved into experimental positions from the static trial [23, 24].

Once the model is appropriately scaled, trials of actual movement can begin to be processed. The first step to analyzing movement is to run the Inverse Kinematics tool to describe the model's joint angles over the time frame of the trial. Discrepancies in the motion can occur either as marker or coordinate errors. Marker errors refer to differences between experimental and model-predicted location. Coordinate error, conversely, deals with one of the model's coordinates (e.g., knee flexion) and how this prescribed value compares to experimental data. For each frame of data, the Inverse Kinematics tool minimizes both the marker and coordinate error using Eq. 1.

$$\min_{q}\left[\sum_{i\in\text{markers}} w_i\left\|\boldsymbol{x}_i^{\text{exp}}-\boldsymbol{x}_i(\boldsymbol{q})\right\|^2+\sum_{j\in\text{unprescribed coords}}\omega_j\left(q_j^{\text{exp}}-q_j\right)^2\right] \qquad (1)$$

In this standard least-squares problem, **q** represents the collected vector of coordinates within the model. q_j^{exp} refers to the calculated experimental value of the coordinate per frame. $\mathbf{x}_i^{exp}$ similarly refers to the experimental position of an individual marker, and is a vector containing position in three dimensions. $\mathbf{x}_i(\mathbf{q})$ is the model marker position and is a function of the model's coordinates. This modified optimization allows for user-specific weighting of specific marker or coordinate errors (w_i and ω_i, respectively) [23, 24]. In this study, equal weighting is used for all coordinates. Finally, some coordinates may add unnecessary error and are prescribed ahead of time. One example is the metatarsophalangeal joint, which is prescribed at 0° of flexion for RFSR trials. These coordinates are therefore omitted from the optimization.

For the gait models used in this project, there is a disparity between the number of degrees of freedom of the model and the joint actuators that are present as a results of model components. OpenSim presents the six degrees of freedom between the model's pelvis and the ground as a joint with three torque and three force actuators, termed "residual actuators" [23, 24]. With the addition of these residual actuators there exists an actuator for every degree of freedom. Dynamic inconsistency will still arise when a model does not completely match the real-life motion of an object. A simplified model like the one used in this study may not contain all the data necessary to account for the totality of GRFs. Noise or other errors from motion capture may also contribute. An especially significant aspect of honing the computational model to reflect real-life behavior is reducing these residual forces and torques to assuage dynamic inconsistency. The familiar equation to satisfy is Newton's Second Law (in the general form of Eq. 2) which relates forces F to the mass m and accelerations a. These are vector quantities as well.

$$\vec{F} + \vec{F}_{\text{residual}} = m\vec{a} \tag{2}$$

For the model to be accurate, the residual force term $F_{residual}$ should be as small as possible. The Residual Reduction Algorithm works to accomplish this by adjusting the mass center of the model and recommending adjustments to model segment masses and kinematic results [23, 24]. The algorithm uses coordinate data from Inverse Kinematic results. Frame by frame, a combination of PD control and static optimization incorporates GRFs and kinematics to describe specific coordinate actuator activity to move the model from state to state in a dynamically consistent fashion [25].

There are no experimental data in this set to inform the position of the torso, so the most accurate assumption is for the mass of the participant's torso to be applied at the center of mass of the pelvis. In this configuration the Residual Reduction Algorithm can run and produce results that are within acceptable parameters [23, 24].

Before the specific knee joint can be analyzed, the contributions to contact force by individual muscles must be known. The model's motion is completely understood from the kinematic results of previous tools: position, velocity, and acceleration, but the forces are generalized by coordinate. Disaggregating the coordinate torques into individual muscle forces is another optimization problem. The Static Optimization tool minimizes the objective function in Eq. 3, subject to the constraints of Eq. 4.

$$J = \sum_{m=1}^{n} (a_m)^2 \tag{3}$$

$$\sum_{m=1}^{n} [a_m f(F_m^0, l_m, v_m)] r_{m,j} = \tau_j \tag{4}$$

where n is the number of muscles in the model; a_m is the activation of muscle m at a discrete time step, F^0_m is its maximum isometric force, $r_{m,j}$ is its moment arm about the j^{th} joint axis; τ_j is the generalized force about the j^{th} joint [23, 24]. The function $f(F_m^0, l_m, v_m)$ encompasses the force-length-velocity relationships of the m^{th} muscle outlined by Thelen [22].

Because compressive force on a joint is not merely the result of ground reaction force; muscle contributions to joint force are far from trivial [26], and specifically, tibiofemoral force has been shown to reduce with altered muscle activations [27], the Static Optimization profile is essential for obtaining accurate results.

This tool minimizes the activation of any specific muscle to ensure an equitable balance between all the model's actuators. Therefore, Static Optimization will preference solutions where more muscles are activated at lower levels. Co-contraction involving more muscles than is necessary is a common output and may lead to increased compressive joint loading [27].

With individual muscle activation calculated for each state, the corresponding force is determined from the muscle's properties. These forces are central to the final step in the process: the Joint Reaction analysis. This tool determines the resultant forces and moments at a certain joint as arising from a collection of known forces, present as both GRFs and muscle contributions, as well as the known kinematics of the joint in a particular frame. OpenSim models consist of rigid bodies linked in kinematic chains. The chain representing the lower extremity can be seen in Fig. 4. A recursive operation allows for joint force balances across the model calculated from the most distal joint to the most proximal [28]. This is accomplished by first performing the force balance given by Eq. 5 centered on the body frame (for the knee joint, this body is the tibia). The linear and angular accelerations of the body from previous steps are represented as $\vec{a}_i$, while the GRFs and muscle forces are $\vec{F}_{external}$ and $\vec{F}_{muscles}$, respectively.

$$\vec{R}_o = \begin{bmatrix} \vec{\tau}_o \\ \vec{F}_o \end{bmatrix} = \boldsymbol{M}_i(\vec{q})\vec{a}_i + \vec{F}_{constraint} - \left(\sum \vec{F}_{external} + \sum \vec{F}_{muscles} + \vec{R}_{i+1}\right) \tag{5}$$

$\vec{R}_o$ is the joint force and moment at the body origin, $\boldsymbol{M}_i(\vec{q})$ represents the mass matrix for the body as a function of the coordinates of the body segment, and $\vec{F}_{constraint}$ accounts for additional constraint forces applied to the body. The recursive nature of this algorithm provides the joint reaction from the more distal body ($\vec{R}_{i+1}$) to contribute to the force balance. In this specific case, the distal ankle joint reaction force is applied to the tibia. Once the requisite force and moment are calculated expressed at the body origin, the actual joint reaction ($\vec{R}_i$) is determined by shifting the force and moment to the joint center [28]. This relationship is given by Eq. 6, where $\vec{r}$ is a vector directed between the body origin and the joint center.

$$\vec{R}_i = \begin{bmatrix} \vec{\tau}_i \\ \vec{F}_i \end{bmatrix} := \begin{bmatrix} \vec{\tau}_o \\ \vec{F}_o \end{bmatrix} - \begin{bmatrix} \vec{r} \times \vec{F}_o \\ \vec{0} \end{bmatrix} \tag{6}$$

The output of this tool is all six of the reaction forces and moments. The compressive loading in the superior-inferior direction was selected for analysis.

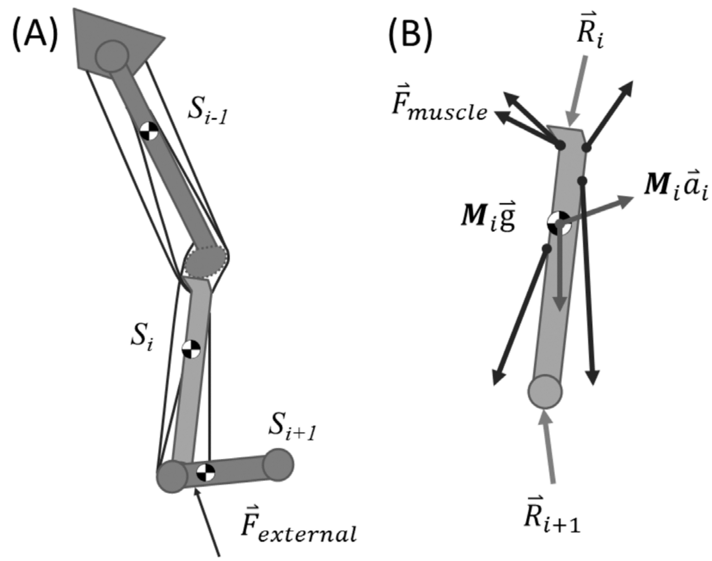

Fig. 4. Kinematic chain representing the lower extremity (A) and force balance on tibia (B). From Steele et. al [28].

Statistical Analysis

Joint reaction data were calculated for 23 participants for each of the 5 trials of RFS and 5 trials of FFS running. Trials that were not successfully processed in OpenSim were not included in the statistical analysis. After excluding these trials, 109 trials of RFS and 105 trials of FFS were compared statistically. Mean and standard error joint reaction data were plotted for comparison. Filtered GRF data were also average across trials and plotted. To quantify the impact transient, the joint reaction force was integrated over the first 15% of stance phase. Impact transient, maximum joint force, and GRF averages were compared using two-sample t-tests (α=0.05).

Results

Significant differences were observed in reaction force using both prescribed metrics. A comparison of the FFS and RFS patterns can be seen in Fig. 5. Large variations in force are observed, particularly in FFS. The average peak compressive tibiofemoral contact force during stance phase was 10.06±1.54 %BW (mean ± s.d.) during RFS and 10.79±1.92 %BW for FFS. The 7.2% increase in FFS joint reaction force was found to be significant ($p < 0.01$). This corresponds with a 4.8% increase in measured vertical GRF

(Fig. 6). Average peak GRF was found to be 2.46±0.24 %BW in RFS and 2.58±0.25 %BW in FFS. This increase in GRF was also found to be significant ($p < 0.001$). Average impact transient for RFS was found to be 0.097±0.020 %BW·s. FFS produced an average of 0.076±0.013 %BW·s, a 21% reduction. During initial contact, impact transient was significantly reduced in FFS running ($p < 0.01$). The impact transient is clearly visible in the plots of both joint and ground reaction force in the RFS condition, and the reduction is equally apparent on the FFS curve.

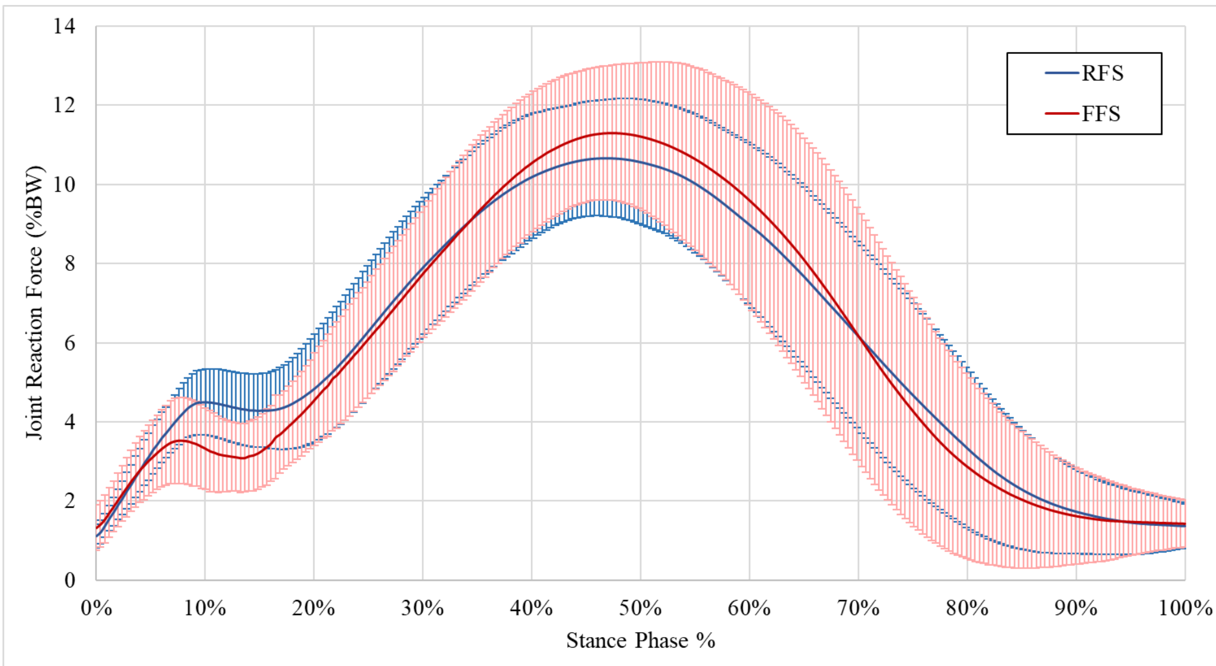

Fig. 5. Compressive tibiofemoral force during stance phase

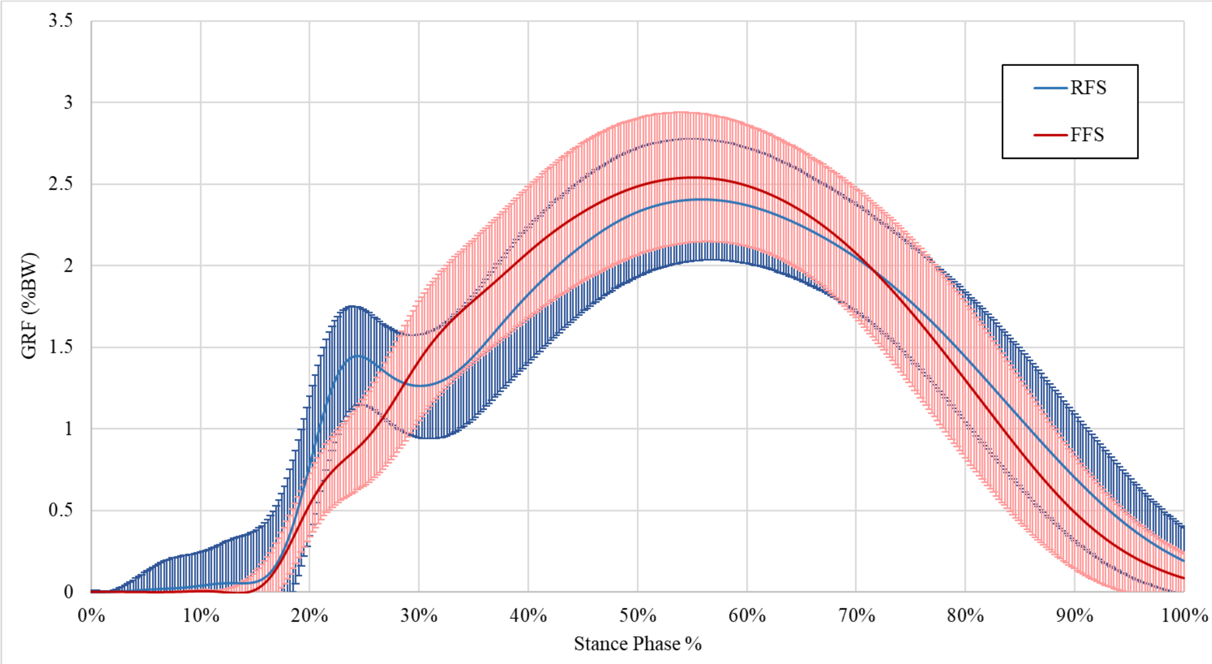

Fig. 6. GRF during stance phase

Discussion

The purpose of this study was to model and analyze the kinetic differences experienced by the knee joint as a result of foot strike pattern alteration. Results indicate that while peak GRFs and total compressive force is increased during FFS running, the impact transient is greatly reduced. This supports previous research that indicated that a FFS strike pattern has the potential to reduce injuries by eliminating this impulse in GRFs [17]. This study was able to correlate this externally-measured impulse elimination to a similar effect in internal loading. Given the efficacy of FFS in reducing this impact, it may be helpful to runners prone to or suffering from knee pain to adopt an FFS pattern. The increase in maximum compressive tibiofemoral force is an unexpected result, as it does not agree with previous research conducted by Rooney et. al that differences between conditions would not be significant for habitual RFS runners [29]. However, the similar increase in GRF in switching to FFS contributes to this difference and is similar to the results found by Boyer et. al [30] Additional compression is created by muscles crossing the knee joint. Specifically, both the medial and lateral gastrocnemius experienced higher activation during FFS (Fig. 7). These results are consistent with previous findings that plantarflexors are more active during FFS [16]. The gastrocnemius contributes to the kinematics of the foot, especially during toe off, which adds to the total compressive load experienced by the knee.

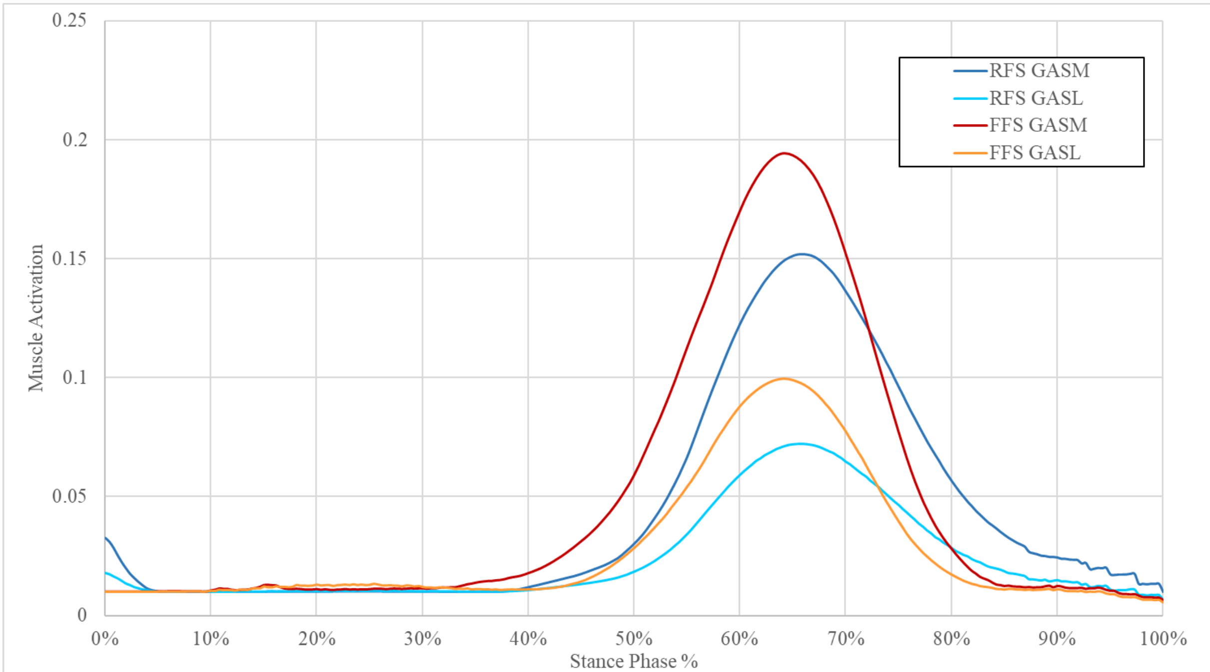

Fig. 7. Medial and lateral gastrocnemius activations during stance phase

This study is limited primarily by the modeling techniques used to predict experimental qualities. Though modeling is a common tool in biomechanics, the particular difficulty of measuring *in vivo* loading makes experimental validation of calculated joint reaction forces impossible. As the product of fallible algorithms, the model may not always present an accurate assessment of internal loads.

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