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Research Article

THE EFFECT OF MEDIAL HAMSTRING WEAKNESS ON KNEE CONTACT FORCES DURING RUNNING

A. Schmitz

Department of Engineering and Technology, University of Wisconsin-Stout

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ABSTRACT

When used to reconstruct the anterior cruciate ligament (ACL), the medial hamstrings graft has been shown to produce lower rates of osteoarthritis (OA) than the patellar tendon graft. The goal of this study was to determine how reducing medial hamstring strength during surgery affects joint contact forces during loading, and hence the joint's proclivity towards OA. A previously developed model of the entire body was used to perform a muscle-actuated forward dynamics simulation of running for two cases: normal muscle strength and medial hamstrings (i.e. semitendinosus and gracilis) weakened by 30%. The muscle forces from these simulations were then used to actuate a discrete element model in a forward dynamics simulation. Weakening the medial hamstrings caused an overall decrease in total hamstrings force by <1%, total quadriceps force by 2%, and cartilage contact force by 4%. This decreased force may be protective against long-term OA and hence may help explain the lower rates of OA in patients who receive medial hamstring grafts.

Keywords: ACL, knee model, medial hamstrings, joint contact loads, serial approach.

Author Correspondence, e-mail: schmitzann@uwstout.edu

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1. INTRODUCTION

Over 200,000 anterior cruciate ligament (ACL) reconstructions are performed every year in the United States [1], which leads to $15,000 in health care costs per procedure [2]. After ACL surgery, knee osteoarthritis (OA) develops within 10 – 20 years in 50% of these knees [3]. In an effort to reduce these long-term complications and associated health care costs, much research has been done on developing programs to prevent ACL injury [4], optimizing surgical techniques [5], and developing effective post-surgical rehabilitation programs [6]. One important surgical parameter to research is the tissue used to replace the ACL.

Typically, ACL reconstructions harvest ligament replacement tissue from two sites in the injured patient: patellar tendon and medial hamstrings [7]. For the patellar tendon graft, an 8 to 11-mm wide graft is taken from the central third of the patellar tendon with adjoining patellar and tibial bone grafts [8]. This graft provides high strength and stiffness properties with a stiffness 84% of an intact ACL [9]. Therefore, this graft is one option used in young athletes involved in dynamic sports to allow for a quick return to sport [8]. However, donor site morbidity can be an issue, e.g. anterior knee pain, knee extensor weakness. Medial hamstrings grafts are the preferred option in young athletes due to smaller incisions for surgery and lower anterior knee pain, which results in lower donor site morbidity [8], [10]–[13]. These grafts are also more comparible to the stiffness of a native ACL, i.e. 103% of an intact ACL [9]. In this graft option, portions of the semitendinosus and gracilis tendons are harvested, folded in half, and combined to create a 10-mm round ACL replacement [8]. Compared to a patellar tendon graft, patients receiving a medial hamstrings graft have reported a better ability to walk on their knees after 2-years post-surgery [14], i.e. walking while kneeling on the ground [15]. However, these studies are typically observational in nature, which precludes a mechanistic explanation of how the medial hamstrings graft affects joint health.

Joint health has been quantified as joint mechanics measures, particularly cartilage contact forces. These forces have been associated with osteoarthritis progression [16]. Specifically, increased cartilage contact forces are positively related to increased rate of cartilage degradation in an osteoarthritic joint [16]. Medial hamstrings grafts have been shown to result in a lower rate of OA than patellar tendon grafts [17]. The goal of this study is to determine how weakening the medial hamstrings affects joint health and help explain the lower OA rate in these patients. Hence, medial hamstring weakness (which results when harvesting the semitendinosus and gracilis tendons for reconstruction [18]) is hypothesized to cause decreased cartilage contact forces during running. Running is an important exercise to investigate because 82% of athletes who receive an ACL reconstruction return to some level of dynamic sporting activities [19].

2. METHODS

Joint contact forces have only been measured *in vivo* using instrumented knee replacements [20]. This method is not available in patients with intact cartilage. Therefore, computational models have been used to study soft tissue loads [20]–[23]. To quantify cartilage contact loads, an open-source musculoskeletal model of the body [24] and discrete element knee model [25] were utilized in a serial approach (Fig.1) [23], [26].

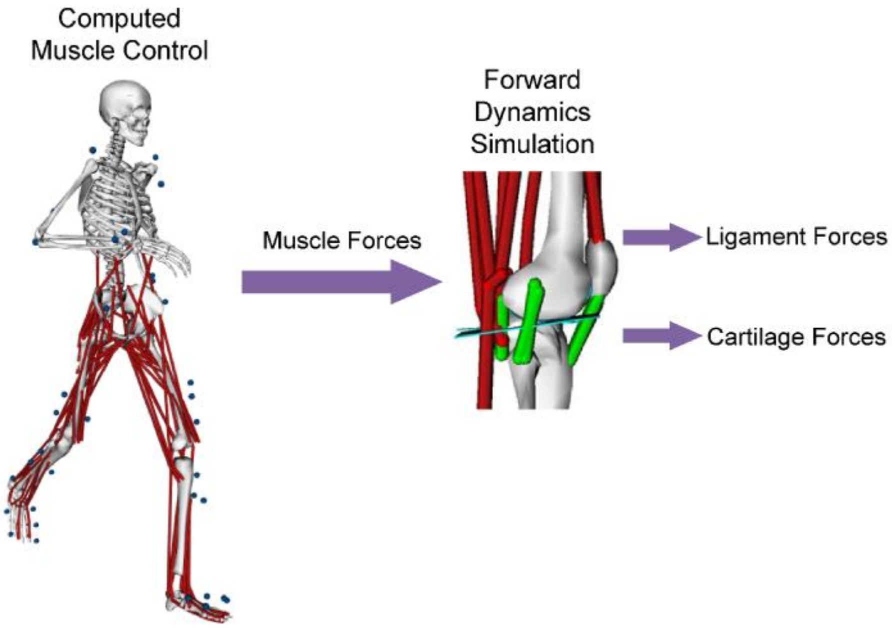


Fig.1. Serial approach where multibody dynamics are first solved using computed muscle control. The resulting muscle forces are used to actuate a discrete element knee model, where soft tissue loads are calculated

First, a previously developed model of the entire body was used to perform a muscle-actuated forward dynamics simulation of running at a self-selected speed of 3.96 m/s [24]. The model was comprised of 12 links representing bony segments of the body and 29 degrees of freedom (dof) for the joints. Each lower extremity contained 5 dof: ball-and-socket hip joints (3 dof), 1 dof custom knee joints, and revolute ankle joints (1 dof). Lumbar-pelvis motion was modeled as a ball-and-socket joint. Each arm was composed of 5 dof: ball-and-socket shoulder joints (3dof), hinge elbow joint (1 dof), and a revolute forearm joints (1 dof). Muscular structures were included as 92 musculotendon actuators and the arms moved using torque actuators at the shoulders. The joint kinematics, kinetics, and muscle activations of this model were validated against experimental data (e.g. electromyography) for running [24]. This model, along with experimental running kinematics and kinetics freely provided by [24], were input into computed muscle control (CMC) [27].

The CMC analysis is composed of three components: proportional-derivative (PD) controller, optimization, and forward dynamics (Fig.2). CMC calculates the muscle forces needed to actuate the model towards experimental kinematics and kinetics. The PD controller is used to determine if the predicted model motion is ahead or behind the experimental data and adjust accordingly. For example, if the model is ahead of the experimental data, the PD controller will decelerate the model to slow down. The PD controller ultimately calculates joint accelerations. The next component, optimization, is used to determine how to actuate the muscles to achieve these joint accelerations. Optimization is needed to reduce muscle redundancy, i.e. number of muscles is greater than the number of degrees of freedom. These muscle forces are then used in a short forward dynamics simulation to actuate the model forward by one time step. This new model location is compared to the experimental data, where the PD controller starts anew.

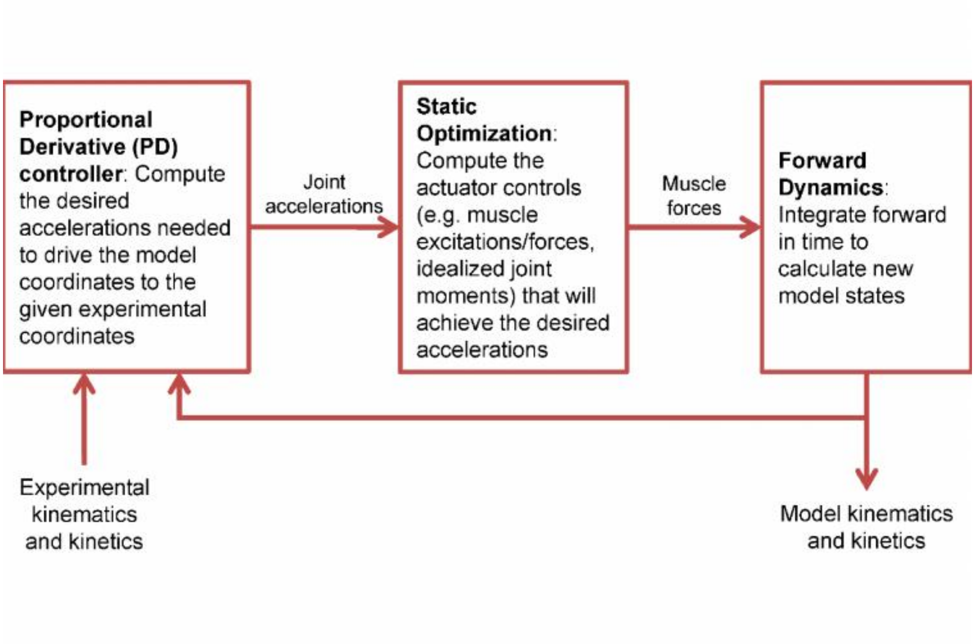


Fig.2. Computed muscle control (CMC) analysis used to calculate muscle forces needed to actuate a model to match experimentally measured kinematics and kinetics

This muscle-actuated forward dynamics simulation of running (i.e. CMC) was performed for two cases: normal muscle strength and weakened medial hamstrings. Muscle weakness was quantified using the maximum isometric force property of the muscle, which was modeled as a Hill-type musculotendon actuator [28]–[30]. For the weakened hamstring model, the maximum isometric force of the semitendinosus and gracilis were each decreased by 30% [31] (Fig.3). This percentage was chosen based on morphological studies done post-surgery. Also, these specific muscles were chosen as they are most commonly used for the medial hamstrings ACL graft [7], [8]. The variables of interest extracted from these CMC simulations were force in the medial hamstrings (semitendinosus and gracilis), lateral hamstrings (biceps long head and short head), semimebranosus, and quadriceps (vastus lateralis, vastus intermedius, vastus medialis, and rectus femoris). Total hamstring force was quantified as the sum of the forces produced by the semitendinosus, semimembranosus, gracilis, biceps femoris long head, and biceps femoris short head. Total quadriceps force was computed similarly as the sum of forces produced by the vastus medialis, vastus lateralis, vastus intermedius, and rectus femoris. The forces in all 92 muscles were subsequently used to actuate a discrete element knee model.

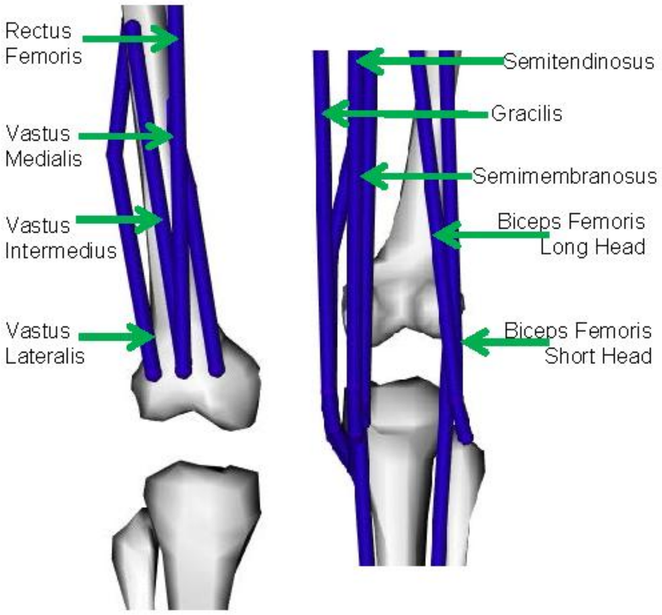


Fig.3. Muscles of interest. The medial hamstrings were defined as the semitendinosus and gracilis since these muscles are used for ACL reconstructions. The biceps femoris long and short head were considered the lateral hamstrings

The muscle forces from CMC were used to actuate a previously developed discrete element model [25] using a forward dynamics simulation. The discrete element knee model consisted of a 6 dof tibiofemoral joint and 1 dof patellofemoral joint. Knee motion was constrained via 18 non-linear elastic ligaments and contact. Contact was quantified with an elastic foundation model between the medial and lateral tibial plateaus and the femoral condyles. The geometry of the tibia was assumed to be planar and the femur geometry taken from MRI data (as described in a previous study [25]). The muscles included in the discrete element knee model were the same as those from the whole body model [24].

Forward dynamics was used actuate the discrete element knee model with the CMC results of the whole body model. The discrete element knee model was modified for the two cases: normal muscle strength and weakened medial hamstrings. The variables of interest extracted from these forward dynamics simulations results were force magnitude in the medial tibiofemoral compartment and lateral femoral compartment, which were computed using the elastic foundation contact model. Muscle and cartilage contact forces were compared to test the study hypothesis: medial hamstring weakness will cause decreased cartilage contact forces during running. Since these simulations were run for a single subject [24] more detailed statistical measures, such as minimal detectable change [32], were not assessed.

3. RESULTS

Decreasing the gracilis and semitendiosus strength by 30% each resulted in a decreased peak force for the gracilis and semitendinosus by 44% and 42%, respectively (Table 1 and Fig.4). In contrast, an increased peak force was experienced in the semimembranosus by 5%, biceps femoris long head by 2%, and biceps femoris short head by <1%. Ultimately, weakening the medial hamstrings caused an overall decrease in total hamstrings force by <1%. Peak force was also decreased for all of the quadriceps muscles by 1 – 3%, with a decrease of 2% for the total quadriceps force (Table 2 and Fig.4). The maximum forces in the medial and lateral tibiofemoral cartilage were decreased by 1% and increased by 3%, respectively, after the medial hamstrings were weakened (Table 3 and Fig.5). This resulted in an overall 4% decrease of the total tibiofemoral contact force.

Table 1 Change in Maximum Hamstrings Force

	Semitendinosus (N)	Semimembranosus (N)	Gracilis (N)	Biceps Femoris Long Head (N)	Biceps Femoris Short Head (N)	Total Hamstrings Force (N)
Normal	221	601	120	493	471	1839
Weak	156	633	83	500	473	1830
% Change	-42	5	-44	2	0	0

Table 2 Change in Maximum Quadricpes Force

	Vastus Medialis (N)	Vastus Lateralis (N)	Vastus Intermedius (N)	Rectus Femoris (N)	Total Quadriceps Force (N)
Normal	999	2038	1194	998	4268
Weak	968	2022	1157	989	4185
% Change	-3	-1	-3	-1	-2

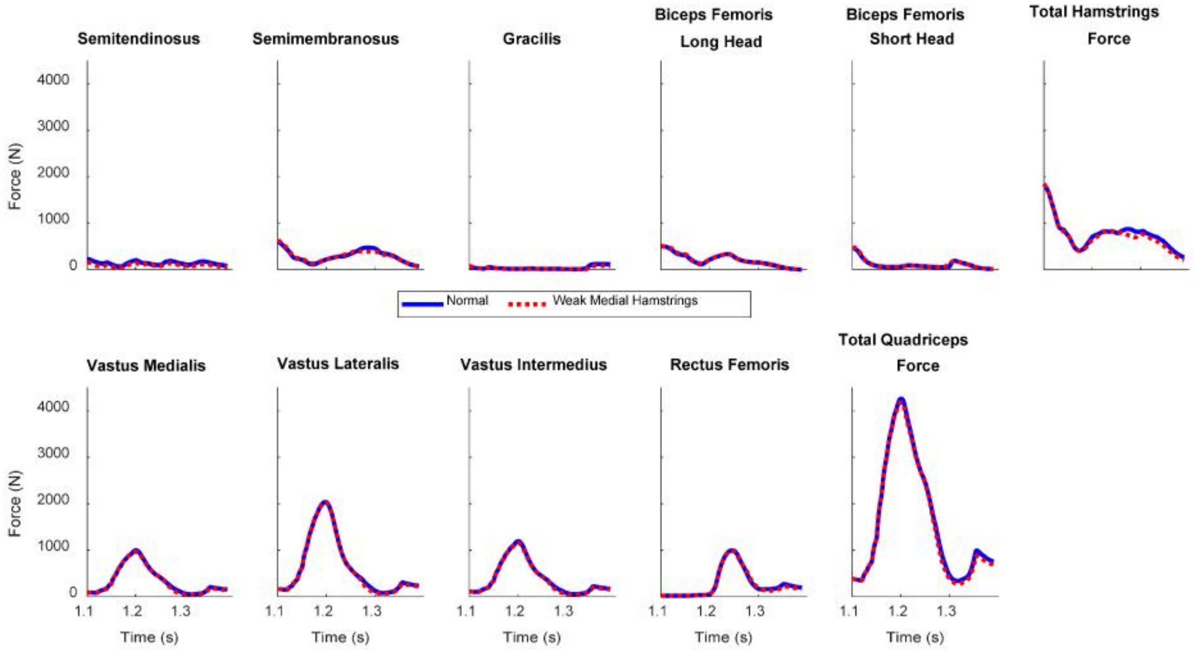


Fig.4. Effect of medial hamstring weakness on quadriceps and hamstrings forces. The curves are shown for the stance phase of running for a single subject

Table 3 Change in Maximum Cartilage Contact Force

	Medial (N)	Lateral (N)	Total Contact Force (N)
Normal	3612	2999	5450
Weak	3589	3102	5227
% Change	-1	3	-4

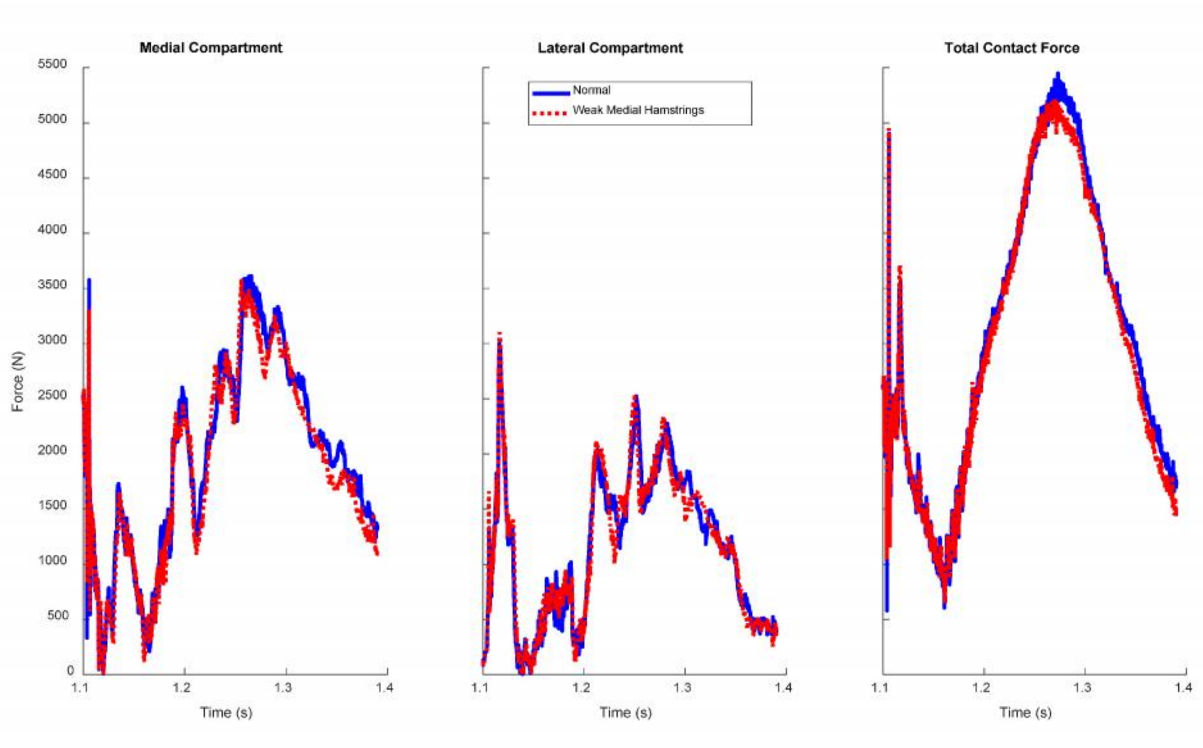
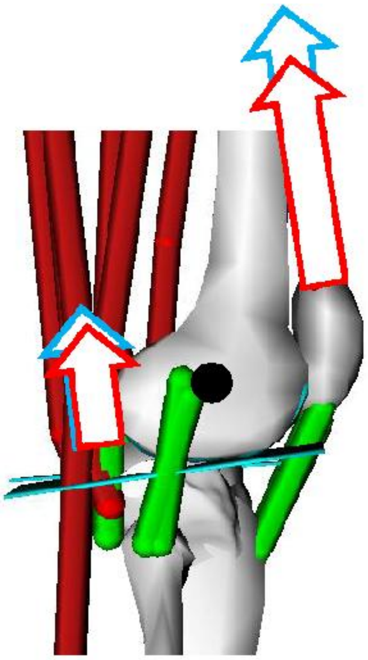


Fig.5. Effect of medial hamstring weakness on cartilage contact forces. The curves are shown for the stance phase of running for a single subject

4. DISCUSSION

In ACL reconstruction surgery, graft tissue is harvested from the medial hamstrings, specifically the semitendinosus and gracilis. The medial hamstrings graft is preferred over patellar tendon grafts since they have shown a lower rate of OA occurrence [17]. The goal of this study was to develop a mechanistic explanation for how a medial hamstrings graft would affect joint health. Muscle-actuated forward dynamics simulations of running were used to calculate muscle forces and joint contact loads for a healthy knee and a model with decreased medial hamstrings strength.

Ultimately, weakening the medial hamstrings caused an overall decrease in total hamstrings force by <1% and total quadriceps force by 2%. This agrees with other studies that have shown quadriceps weakness and dysfunction in those with ACL reconstructions [33]. This uneven decrease in force between the hamstrings and quadriceps can be explained using a free body diagram of a sagittal view of the knee (Figure 6) and performing a sensitivity analysis.



$$I\ddot{O} = F_{quad} * r_{quad} - F_{hamstrings} * r_{hamstrings}$$

Fig.6. When the medial hamstrings are weakened, the quadriceps are more affected than the hamstrings. This is due to the change in moment arms that result as the joint slides in the anterior/posterior direction

A change in a torque about the knee joint center is due to a change in both the quadriceps and hamstrings torques (Eq. 1).

$$\delta T = \delta T_{quad} - \delta T_{hamstrings} \qquad \text{Eq. 1}$$

These change in torques are caused by changes in the muscle forces and moment arms (Eq. 2).

$$\delta T = [\delta F_{quad} * r_{quad} + F_{quad} * \delta r_{quad}] - [\delta F_{hamstrings} * r_{hamstrings} + F_{hamstrings} * \delta r_{hamstrings}] \qquad \text{Eq. 2}$$

In both the normal and weakened simulations, the same amount of sagittal plane torque to flex and extend the knee was produced. Therefore, the change in quadriceps torque will equal the change in hamstrings torque (Eq. 3).

$$[\delta F_{quad} * r_{quad} + F_{quad} * \delta r_{quad}] = [\delta F_{hamstrings} * r_{hamstrings} + F_{hamstrings} * \delta r_{hamstrings}] \quad \text{Eq. 3}$$

As the quadriceps force decreases, the tibia will decrease in anterior translation and increase the moment arm of the quadriceps. This will decrease the hamstrings moment arm by the same amount. This change in quadriceps moment arm has been shown to vary by around 2 mm [34]. Also, the average quadriceps moment arm has been shown to be 4 cm and the hamstrings around 2 cm [35]. Substituting these values into Eq. 3, along with the values of 4200 N for the force in the quadriceps and 1800 N for the force in the hamstrings (Table 1 and 2), gives Eq. 4.

$$\delta F_{quad} = -300 + 0.5 * \delta F_{hamstrings} \quad \text{Eq. 4}$$

This equation shows that the change in quadriceps force is greater than the change induced in the hamstrings, which is what is shown in the results. This equation also shows both forces decrease together, as opposed to one increase and one decrease, to create a net zero change in the sum of torques.

The decrease in muscle forces resulted in a total decrease in cartilage contact force by 4%. Since increased force has been shown to expedite cartilage degeneration, this decreased cartilage load may help explain the low rates of OA in those with a medial hamstring graft [17]. Although, some level of mechanical loading is needed for cartilage growth and remodeling [36]. Therefore, more work is needed to better elucidate the balance between too little loading that can cause atrophy versus too much loading that can cause degeneration.

When interpreting the results of this study, there are some study design factors to consider. First, CMC uses an optimization strategy to solve for muscle forces in a redundant system. This optimizer activates muscles to use the least energy possible, which is the sum of volume times muscle activations squared [27]. It is unclear if this neuromuscular strategy is utilized after an ACL surgery and how this varies between subjects. Although, this control strategy has been used for abnormal gait patterns in other studies investigating muscle weakness [30]. A different control strategy would likely change the values of the results in the current study; although, the patterns of change would likely remain the same, i.e. medial hamstrings weakness leading to decrease joint contact loading, as muscle coordination patterns and timely are fairly consistent in the presence of muscle weakness [30]. Second, the model was assumed to run at the same speed with the same kinematics and kinetics both in the normal and weakened states. This was done to isolate the effect of muscle weakness on contact forces rather than confounding the effect by adding running speed as a variable.

5. CONCLUSION

In summary, musculoskeletal modeling was used to quantify the distribution of joint contact loads when the medial hamstrings (i.e. semitendinosus and gracilis) are compromised, which is common in ACL reconstructions. Overall, both quadriceps and hamstrings forces decreased. Cartilage contact loads also decreased, which may have implications for joint health and long-term OA development. The next step in this study is to elucidate the optimal loading needed to maintain cartilage health.

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