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The effects of knee support on the sagittal lower-body joint kinematics and kinetics of deep squats

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ABSTRACT

Little work has been done to examine the deep squat position (>130° sagittal knee flexion). In baseball and softball, catchers perform this squat an average of 146 times per nine-inning game. To alleviate some of the stress on their knees caused by this repetitive loading, some catchers wear foam knee supports.
Objectives: This work quantifies the effects of knee support on lower-body joint kinematics and kinetics in the deep squat position.
Methods: Subjects in this study performed the deep squat with no support, foam support, and instrumented support. In order to measure the force through the knee support, instrumented knee supports were designed and fabricated. We then developed an inverse dynamic model to incorporate the support loads. From the model, joint angles and moments were calculated for the three conditions.
Results: With support there is a significant reduction in the sagittal moment at the knee of 43% on the dominant side and 63% on the non-dominant side compared to without support. These reductions are a result of the foam supports carrying approximately 20% of body weight on each side.
Conclusion: Knee support reduces the moment necessary to generate the deep squat position common to baseball catchers. Given the short moment arm of the patella femoral tendon, even small changes in moment can have a large effect in the tibial-femoral contact forces, particularly at deep squat angles. Reducing knee forces may be effective in decreasing incidence of osteochondritis dissecans.

1. Introduction

As of spring 2016 there were 13.46 million baseball and softball players in the United States (Statista, 2017). In baseball, there is a significant body of knowledge studying upper extremities, specifically pitchers' arms (Fleisig et al., 2011, 1995; Lyman et al., 2002; Mirowitz and London, 1992; Olsen Ii et al., 2006), but little has been studied about the lower body and more specifically, catchers. Yet, approximately 1.6 million of the baseball players in the US are catchers (Statista, 2017). While not a contact sport, injury prevention and safety are still major areas of concern for baseball and softball players. Injuries in catchers are commonly seen at the knees.

The repetitive hyperflexion squatting motion performed by catchers has been associated with osteochondritis dissecans (OCD) in the knees (McElroy et al., 2016). OCD is characterized by lesions forming at joints from cartilage that has died due to lack of blood flow (Aichroth, 1971). The bone and cartilage in these lesions can loosen and this causes pain and hinders the movement of the joint. OCD also increases the risk of developing arthritis at that joint (Aichroth, 1971; McElroy et al., 2016). When compared to other positions, catchers develop OCD at a younger age and specifically in the posterior femoral condyle (McElroy et al., 2016). In his study, McElroy described "Catcher's Knee" to be a "posterior femoral condylar OCD lesion seen with the repetitive and persistent hyperflexion seen in catchers."

Little work has been done to quantify the kinetics and kinematics of deep squats. Studies thus far have primarily focused on what happens to the knee up to 90° (Han et al., 2013), but catchers typically go well past this mark. In 1986, Nisell and Eckholm looked at the muscle forces required during a parallel squat to rupture the patellar tendon of a powerlifter. They found the force through the patellar tendon at the time of rupture was 17.5 times body weight. They also found that rupture occurred during the beginning of the ascent from the squat, suggesting that the peak force experienced by the knee occurs as the squatter's muscles fire to

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bring their body back to the upright position. This motion is performed by catchers on nearly every play.

A cadaver model has been used to demonstrate that deep knee flexion results in forces high enough to damage cartilage (Thambyah et al., 2005). A single stress over 25 MPa can cause damage to cartilage; however, cyclic stress can cause damage to cartilage at 5–10 MPa (Farquhar et al., 1996). The simulated deep squats performed by Thambyah on cadavers found a stress of 26.6 MPa across the knee. This is past the damage limit for cartilage as well as an 80% increase from the 14.1 MPa the knee experienced during simulated walking. Additionally, it has been estimated that the maximum in vivo contact forces in the knee are 6.5 times bodyweight at 90° flexion, but it has been hypothesized that you could expect twice this in trained athletes (Huberti and Hayes, 1984). In the deep squat position, it is possible that there will be contact between the thigh and the calf. Zelle et al. (2007) measured this contact and found that contact occurs starting at approximately 135° and reaches a maximum of 34.2 %BW at 152°. To arrive at these conclusions the authors used a pressure mat which found the resultant contact force, but not the direction in which this force acts. Also, the authors did not correlate this contact force to knee moment or a reduction in force across the knee.

McElroy suggests that prevention of what he has termed "Catcher's Knee" (OCD in the posterior femoral condyle) "may be influenced by "catch counts" and equipment such as triangular foam "knee savers" that are placed behind the knee to reduce knee flexion or unload the knee in hyperflexion." However, he was not able to quantify the effects of wearing the foam supports. These foam supports (Fig. 1D) are used by catchers from little league up through Major League Baseball. While its widespread use would suggest that foam knee support is helpful to baseball catchers, it is not known how this support helps catchers. McElroy's study was a retrospective review of records and consequently he could not examine the effects of knee supports on the knee. The study presented here develops a method to do just that.

This work develops instrumentation and computational musculoskeletal models to determine the loads experienced by the knee supports in the deep squat position. These devices are used to collect experimental data to inform the computational model of human squatting that allows for the calculation of joint moments in deep squats (>130° sagittal knee flexion). Ultimately, this work provides a method for and quantifies the lower body kinematics and kinetics in a deep squat, with and without knee support, to better understand the effects of the deep squat position on the joints of the lower body.

2. Methods

Experimental data was collected for ten subjects, two females and eight males, averaging 17.5 ± 5.9 years of age, 179.5 ± 8.8 cm in height, and 74.61 ± 9.59 kg in mass. All tests were conducted in the Motion Analysis and Motor Performance Laboratory at the University of Virginia. Subject consent was approved by the University of Virginia's Human Investigation Committee and was obtained for all subjects.

Subjects performed deep squats (>130° sagittal knee flexion) under three conditions: without knee support (Condition 1) (Fig. 1A), wearing off-the-shelf foam knee support (Condition 2) (Fig. 1B), and wearing custom-built, instrumented knee support (Condition 3) (Fig. 1C).

For each trial, subjects were instructed to step forward onto two force plates, one under each foot, squat to assume the position of a baseball catcher (>130° sagittal knee flexion), hold this position for two seconds, and then stand, pausing briefly before squatting and after standing.

3. Data collection

Three-dimensional kinematic data were collected using an eight camera Vicon Motion Analysis System (Oxford Metrics, UK) at 120 Hz, and a modified full-body Plug-in-Gait marker set. This marker set included 35 markers and substituted the left and right posterior superior iliac spine (LPSI and RPSI) markers with a triangular cluster of three markers over the sacrum. The cluster over the sacrum

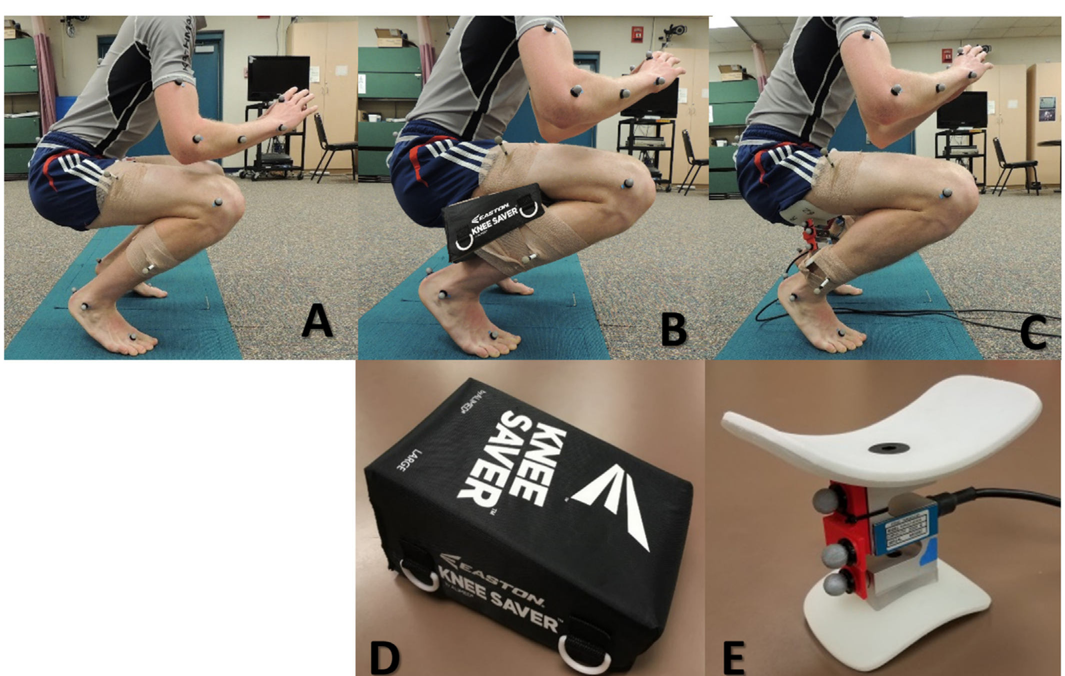

Fig. 1. Three conditions for which data was collected. (A) is a subject wearing no knee support. (B) is a subject wearing the foam knee support. (C) is a subject wearing the instrumented knee support. (D) is a detail of the foam support used. (E) is a detail of the instrumented knee support developed to capture the force through the support in the deep squat position.

was implemented to facilitate the use of virtual markers for the left and right anterior superior iliac spine (LASI and RASI), as these markers tend to be occluded during the deep-squat.

Ground reaction force data were collected at the feet using two in-ground force plates (Kistler; Winterthur, Switzerland. Bertec; Columbus, OH, USA). Data for the force experienced by the knee support was captured using an instrumented knee support. This custom-built instrumented support (Fig. 1E) consists of a one degree of freedom load cell (Interface, Scottsdale, AZ, USA) with a custom interface to allow the subject to squat directly onto the load cell (Fig. 1C). This surrogate was also developed to mimic the size and shape of the commercially available knee supports. Attached to each support is a triad of markers to give the 3D orientation and position of each support through every trial. The instrumented supports were worn at the same location as current foam knee support devices (posterior mid-calf). For each trial, data was truncated to include just before the subject began to descend into the squat until just after the subject had returned to a fully upright position.

4. Models

This work was part of a larger study, so all data were collected for a three-dimensional, 19 segment, 18 joint, 44 DoF, human-body model and reparametrized for each subject from individual anthropometric data (age, weight, height, and gender), in MSC.Adams, using the LifeMod plug-in (Biomechanics Research Group; San Clemente, CA). The 19 model segments included the: head, neck, upper torso, central torso, lower torso, clavicle (2), upper arms (2), lower arms (2), hands (2), upper legs (2), lower legs (2), and feet (2). The segments' physical properties were defined using the Generator of Body Data (GeBOD) database (Cheng et al., 1994). The 18 joints were each specified as tri-axis hinge joint arrangements, with the following exceptions: the sternoclavicular, elbow, and wrist joints were modeled as two-axis hinge joints, and the knee as a one-axis hinge joint. Measured marker positions were exported from VICON to the LifeMod model, and inverse kinematics were performed to calculate the sagittal joint kinematics of the lower body in each of the three conditions.

Inverse dynamic analyses were performed to determine the moments generated at all 18 modeled joints. The hips, knees, and ankles were then further analyzed for comparison between squat conditions. The position and direction data from the clusters on the load cell savers combined with the analog force data recorded by the cell allowed the force experienced by the knee supports to be incorporated into the model. Specifically, the load cell force was applied along the long axis of the load cell in equal and opposite directions onto the upper leg and lower leg at the point that the force vector from the load cell support crossed each of those bodies (Fig. 2).

Kinetic analysis is not possible for Condition 2 (off-the-shelf knee support) as the loads they supported were not measurable, as noted by McElroy (McElroy et al., 2016). In order to compare the joint moments in the supported squat position with the joint moments of being supported in the support squat position a fourth condition was added during modeling: kinematics of wearing the force measured by the instrumented supports was not applied to the model. This gives the four conditions considered in this work to be: (1) squatting with no device, (2) squatting with the foam device, (3) squatting with the instrumented support where the force measured by instrumented support was applied to the model, and (4) squatting with the instrumented support where the force measured by the instrumented support was not applied to the model. Since there could possibly be small differences between the kinematics between Condition 2 and Condition 3, Condition 4

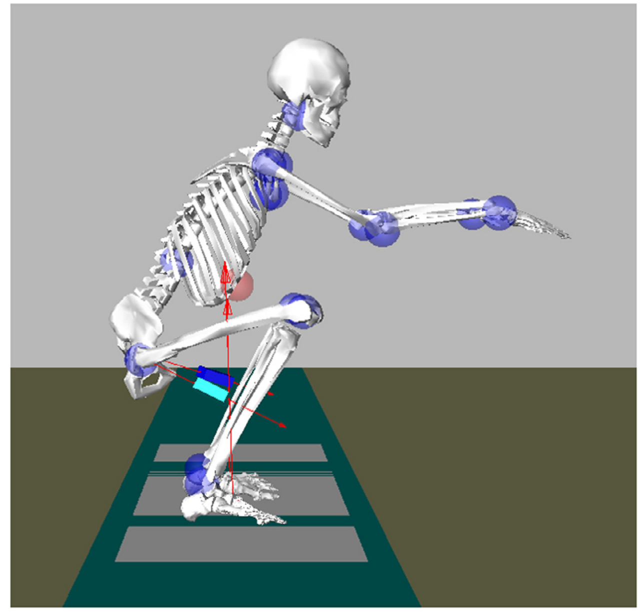

Fig. 2. A completed model rendered in MSC. Adams with the instrumented load cells represented by the two blue boxes floating between the upper and lower legs. The two vertical red arrows denote the ground reaction forces measured by the force plates under each of the subjects' legs. The red arrows extending out of each end of the blue boxes denote the load experienced by the knee support during data collection. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

allowed for a direct comparison of kinetics between knee support and no knee support with both models' kinematics held in a constant position, representative of a supported position.

5. Statistics

A minimum of three squats from each subject under each of the test conditions were averaged to generate the data set for statistical analysis. Analysis was performed on the steady-state position in the deepest part of the squat, shown in the shaded region of the following graphs. Only the sagittal plane was analyzed. Flexion at the joint was assumed to be the positive direction. All kinetic values were normalized, forces by subject weight (%BW) and moments by subject mass and height (Nm/kg-m). Statistical analysis was performed with a one-way repeated-measures ANOVA test, followed by Bonferri correction to further examine the results of the ANOVA test. Significance was determined to be at $p < 0.05$.

6. Results

The model developed for human squatting was successfully used to analyze deep squats. Data showed that knee support dramatically reduces the moment generated by the knee joint to hold the deep squat, while minimally changing the position of the subject (Fig. 3).

No differences in joint kinematics were measured at the hip (Table 1), knee (Table 2), or ankle (Table 3) when comparing no support and foam support. When comparing the no support and instrumented support conditions there was an average of $6.4 \pm 7.9^\circ$ ($p < 0.01$) difference between the mean knee flexion. On average the subjects were sitting with 6.4° less knee flexion using the instrumented support, than when they were with no support.

The load cells in the instrumented knee supports experienced an average force of 18.11 ± 6.42 percent body weight (%BW) per side across all of the subjects. There were no differences between

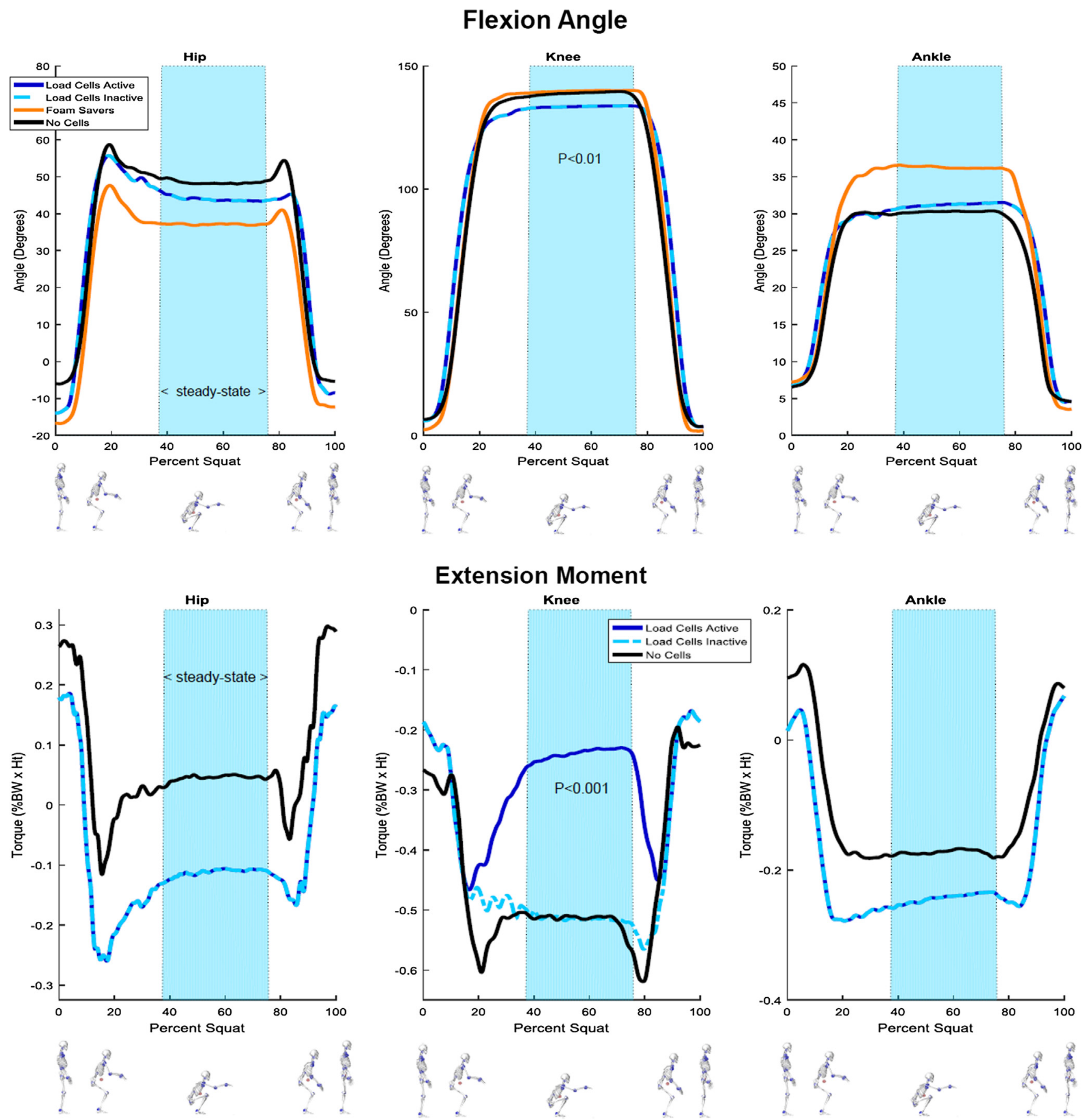

Fig. 3. Plotted results of joint flexion angle and extension moment. Data for this study was analyzed only for the steady-state position, highlighted by the light blue rectangle in each plot, to allow for characterization of deep squat position. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

the loading of the dominant and non-dominant sides, with the dominant side averaging 18.05 ± 6.10 %BW and the non-dominant averaging 18.41 ± 6.74 %BW.

No differences in joint kinetics were measured at the hip (Table 1) or ankle (Table 3) across all three conditions (wearing nothing, wearing the instrumented supports without the support force applied, wearing the instrumented supports with the support force applied). The support condition significantly reduced the sagittal knee moment in the deep squat position (>130° sagittal knee flexion), $p < 0.0001$ (Table 2). This reduction was 43% on the dominant side and 63% on the non-dominant when comparing no support to instrumented support with support forces applied. These reductions are the result of a 31.95 Nm and 38.42 Nm decrease in moment calculated at the knee on the dominant and non-dominant sides respectively.

There was also a significant difference at the knees measured while wearing the instrumented supports comparing the sagittal moments of the support force applied and not applied, $p < 0.0001$. These differences show reductions of 38.14 Nm on the dominant side and 35.28 Nm on the non-dominant side by applying the support force. These reductions are possible by the load cells carrying approximately 20% of body weight on each leg.

Table 1
Hip angle and torque in all conditions. Displayed are the results of model analyses at the hip joint. No significant differences were seen between any of the conditions.

Condition	Measure	Side	Hip ± SD
Nothing	Angle (degrees)	Dominant	59.4 ± 11.9
		Non-dominant	56.2 ± 13.4
	Moment (Nm/kg-m)	Dominant	−0.175 ± 0.122
		Non-dominant	−0.168 ± 0.111
Foam support	Angle (degrees)	Dominant	43.4 ± 16.0
		Non-dominant	40.5 ± 16.1
Load cells (force inactive)	Angle (degrees)	Dominant	53.0 ± 15.5
		Non-dominant	52.3 ± 16.0
	Moment (Nm/kg-m)	Dominant	−0.154 ± 0.091
		Non-dominant	−0.175 ± 0.106
Load cells (force active)	Angle (degrees)	Dominant	53.0 ± 15.5
		Non-dominant	52.3 ± 16.0
	Moment (Nm/kg-m)	Dominant	−0.154 ± 0.091
		Non-dominant	−0.175 ± 0.106

Table 2
Knee angle and torque in all conditions. Displayed are the results of model analyses at the knee joint. Significant differences are denoted with superscript numbers and asterisks, where * denotes $p < 0.01$ and *** denotes $p < 0.0001$.

Condition	Measure	Side	Knee ± SD
^{1}Nothing	Angle^{3*,4*} (degrees)	Dominant	139.9 ± 5.4
		Non-dominant	139.2 ± 6.2
	Moment^{4***} (Nm/kg-m)	Dominant	−0.551 ± 0.141
		Non-dominant	−0.444 ± 0.182
^{2}Foam support	Angle ^{3*,4*} (degrees)	Dominant	139.3 ± 1.3
		Non-dominant	140.4 ± 1.8
^{3}Load cells (force inactive)	Angle^{1*,2*} (degrees)	Dominant	133.3 ± 5.4
		Non-dominant	133.0 ± 4.6
	Moment^{4***} (Nm/kg-m)	Dominant	−0.596 ± 0.101
		Non-dominant	−0.430 ± 0.140
^{4}Load cells (force active)	Angle^{1*,2*} (degrees)	Dominant	133.3 ± 5.4
		Non-dominant	133.0 ± 4.6
	Moment ^{1***,3***} (Nm/kg-m)	Dominant	−0.312 ± 0.069
		Non-dominant	−0.176 ± 0.125

Table 3
Ankle angle and torque in all conditions. Displayed are the results of model analyses at the ankle joint. No significant differences were seen between any of the conditions.

Condition	Measure	Side	Ankle ± SD
Nothing	Angle (degrees)	Dominant	29.9 ± 6.8
		Non-dominant	29.3 ± 6.4
	Moment (Nm/kg-m)	Dominant	-0.202 ± 0.111
		Non-dominant	-0.225 ± 0.135
Foam support	Angle (degrees)	Dominant	35.2 ± 7.2
		Non-dominant	34.3 ± 7.8
Load cells (force inactive)	Angle (degrees)	Dominant	32.1 ± 8.4
		Non-dominant	30.5 ± 7.8
	Moment (Nm/kg-m)	Dominant	-0.227 ± 0.105
		Non-dominant	-0.269 ± 0.128
Load cells (force active)	Angle (degrees)	Dominant	32.1 ± 8.4
		Non-dominant	30.5 ± 7.8
	Moment (Nm/kg-m)	Dominant	-0.227 ± 0.105
		Non-dominant	-0.269 ± 0.128

7. Discussion

The successful development of a model of humans in deep squats allowed us to calculate the joint moments of subjects both while they were wearing knee support and when they were not. The integration of the forces experienced by the instrumented knee support facilitated the comparison between the supported and unsupported conditions. The kinetics of knee support were compared in two ways. First, a direct comparison was made in the instrumented support condition as this position allowed for simulation with the support forces applied and absent. In this case, kinematics are held exactly constant and the only factor contributing to differences in kinetics is the application of the support force. Second, the instrumented support condition with support forces activated could be compared to the no support condition. In this case, kinematics were not held exactly constant. However, this study shows that these conditions do not have any relevant differences in kinematics.

While little data exist on the kinematics and kinetics of deep squats, there is literature reporting data on medium squats (near

90° knee flexion). Our model calculated knee moment at 90° to be −4.385 ± 1.95 Nm/kg-m which is similar to previously reported data on medium squats, −4.9 ± 2.0 Nm/kg-m (Han et al., 2013). This confirmed that our model was producing accurate joint moments. Ours may be slightly lower as it was not a static pose at 90°, but a capture of the dynamic pose on the way down to 130°.

Upon further examination of kinetics we found that the total load experienced by the instrumented support (~20 %BW) was similar to reported data for thigh-calf contact force of 35 %BW (Zelle et al., 2007). However, catchers in the static squat are only experiencing a knee flexion of 133° while wearing the supports, which is only at the very beginning of the thigh-calf contact range according to Zelle et al. (2007). This suggests that thigh-calf contact is minimal in this higher catcher's squat. Also, this suggests that the support allows catchers to stay in a higher squat while experiencing the support that would be provided by sitting on their lower legs in a deeper squat. This is important as the catcher's squat is an active position, where the catcher is not attempting to reach their maximum knee flexion, but rather they are ready to return to standing at a moment's notice. If catchers were squatting deep enough to rely on thigh-calf contact force to relieve stress in their knees, it is unlikely that they would be able to stand-up as quickly. Future work would need to be done to quantify this hypothesis.

The accuracy of the pelvic cluster to predict the ASI's position via virtual markers was confirmed via spot checking three trials to confirm that virtual markers coincided with the real LASI and RASI markers when they were present at the beginning and end of every trial. The root mean squared error was computed in 3 directions and found to be 4.016 mm in the x-direction, 3.955 mm in the y-direction, and 7.0739 mm in the z-direction (vertical). This difference is less than the radius of the markers used for data collection and consequently deemed an acceptable error range for the virtual markers.

When examining the kinematic results, there was a mean difference of 6.4° between the subjects wearing nothing and the subjects wearing the instrumented supports. It does not seem likely that the subject would be able to detect this small change in position. In studies examining other aspects of squatting, the standard deviation on knee flexion in the squat position was never less than 8° (Charlton et al., 2017; Domire and Challis, 2007; Han et al., 2013; Smith et al., 2008), further supporting that variation this small is common within the same activity, so a variation between activities that is equally as small may be statistically significant, but clinically would not be noticeable. This difference in sagittal knee flexion angle is likely an artifact of the instrumented knee supports. The instrumented supports are slightly larger than the foam supports used in this study. Additionally, the instrumented supports are solid and consequently do not exhibit deformation during squatting as seen with foam supports. However, squats were examined in this work in the static position, so compliance of the support does not change the force exerted by the support onto the upper and lower legs. Just as a spring has a constant force in a static position, so do the foam supports under compression.

It is also important to note that while there was a difference in knee flexion angle between wearing nothing and wearing the instrumented knee supports, this difference is not reflected in the knee moments of these conditions. When examining the moments required to hold this position in the two conditions there is also no difference. At more than 130° sagittal knee flexion a 6° difference in knee flexion does not change the length of the moment arm to the center of mass of the body very much. With the femur horizontal, the 6° change had little effect on the moment arm of body weight about the knee.

Regardless of slight position change with the instrumented knee supports, there is a significant reduction in the moment at the knees when the force supported by the knee supports is applied to the lower body. If kinematics are held constant, by comparing the subject wearing the instrumented supports and turning the force application on and off, the reduction in the moment is approximately 50%, across both sides. This is reasonable as we found that each instrumented support is carrying approximately 20% of total body weight. Each leg is carrying approximately 50% of total body weight and each support is taking this ~20 %BW load off of each leg, resulting in the supports taking at least 40% of the load on each side. Specifically, we found this reduction to be 43% on the dominant side and 63% on the non-dominant side. This reduction is similar to a 59% reduction in compressive knee force seen at 155° that occurs due to thigh-calf contact (Zelle et al., 2009).

If we consider the knee moment of 70 Nm found here in the no support condition and a moment arm of the patellar and quadriceps tendon to the center of the knee, being the tibiofemoral contact point, of 4 cm (Erskine et al., 2014; Im et al., 2015; Krevolin et al., 2004), we find the force through the tendons to be 1750 N. Similarly, in the supported condition with a knee moment of 33 Nm found, we find a force of 825 N through the tendons. This immediately suggests that the knee supports reduce 925 N of force in the tendons. Further, assuming the quadriceps and patellar tendons act as a pulley over the knee joint, these tendons exert a resultant force between the patella and femur and, consequently, between the femur and tibia. Specifically, this resultant force across the knee can be defined as $F_R = 2 * \cos\left(\frac{\theta}{2}\right) * F_T$, where F_T is the force in the tendons and θ is the angle between the tendons. At the deep squat position examined in this work, knee flexion was found to be approximately 135°, giving a half angle between the quadriceps and patellar tendon to be 22.5°. Following this formula gives a resultant force of 3233 N across the knee in the unsupported case and 1542 N in the supported case. This is a reduction of 1709 N across the knee. Given that average body weight in this study was 732 N, this is a reduction of 2.34 times body weight while using the support. Based on this cursory observation, it seems that knee support could significantly impact the incidence rate of OCD among baseball catchers. More work is needed to determine if there is a direct relationship between wearing knee support and development of OCD at the posterior femoral condyle.

To fully describe the compressive force at the knee, more studies would need to be done. The next step would be to add the muscles around the knee and the tibial-femoral contact area into the squat model. This data could be collected from a cadaver study to examine contact area and muscle/tendon moment arms about the knee in vitro, and then add these parameters into the model of the knee. This would allow for a more complicated model of the knee to be developed and the calculation of the compressive force at the knee. Further, tracking of the contact area would allow for evidence to show that the area of "Catcher's Knee," as defined by McElroy et al. (2016) is a direct result of microtraumas from the squat position and not from other factors. Future studies could also benefit from improved instrumented knee support design. In the next iteration, the surrogate support could be made with two smaller load cells, one above the other, with foam on either side. With this design, one could account for both the precise size of the existing foam knee supports along with the compressibility of the foam supports.

Conflict of interest statement

The authors have no conflicts of interest to disclose.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2018.10.024.

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