Annals of Biomedical Engineering (2026) 54:181–194
https://doi.org/10.1007/s10439-025-03887-9

BMES BIOMEDICAL ENGINEERING SOCIETY

ORIGINAL ARTICLE

# Biomechanical Analysis of the Effects of Knee Savers on Knee Joint Loading During High Knee-Flexion Tasks

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Received: 24 May 2025 / Accepted: 13 October 2025 / Published online: 24 October 2025


**Abstract**
Occupational activities that require high knee flexion, such as kneeling and squatting, are associated with a higher prevalence of knee osteoarthritis (OA). It is generally accepted that excessive contact pressures in the joint may cause degeneration of healthy knee joints. Ailments in the patellofemoral (PF) joint were a common cause for knee pain in clinical observations. Knee savers have been used in sports and occupational activities to protect the knee joints during tasks involving high knee flexion. The biomechanics of the effects of knee savers on musculoskeletal loading have not been investigated. The purpose of the current study was two-fold: first was to develop a biomechanical model that accounts for the effects of the interface contact forces between thigh and shank, or between thigh-shank and knee savers; and second was to evaluate, using the biomechanical modeling, the effects of knee savers on musculoskeletal loading in the knee joint in high knee-flexion tasks. Five healthy male subjects (age 20.6 ± 1.96 years, body mass 87.74 ± 9.98 kg, height 1.81 ± 0.08 m) participated in the study. The subjects started from a standing posture, squatted down to a working posture, and returned to the standing posture, while subjects' heels remained on the ground during the tasks. The tasks were repeated without and with knee savers. The musculoskeletal loadings in the knees were calculated using inverse dynamic modeling. Our results indicated that the wearing of knee savers in the high knee-flexion tasks helped to reduce the contact forces in the PF joint by about 20%. Our findings suggest that wearing knee savers may help reduce the risks of the knee OA for workers who are frequently required to perform high knee-flexion tasks for extended time.

**Keywords** Soft tissue · Muscle · Knee joint · Knee savers · Inverse dynamics · High knee flexion · Squatting

## Introduction

Osteoarthritis (OA) is a chronic disease that affects 32.5 million people in the US [1]. The prevalence of knee OA among adults 60+ years is approximately 10% (men) and 13% (women) [2]. Epidemiological studies showed that the prevalence of knee OA among workers is associated with their occupational activities. Coggon et al. [3] analyzed the dependence of the knee OA risk on the occupational activity based on the data of 518 knee OA patients who needed surgical treatments; they found that the prevalence of the knee OA among workers who were required to kneel and squat for more than one hour per day was about twice when compared to the average worker. Anderson and Felson [4] found that the radiographic knee OA prevalence for occupations involving high knee bending at ages 55–64 years is about 145% (male) and 249% (female) higher when compared to the average workers. Felson and colleagues [5] reported a 122% increase in radiographic knee OA occurrence in male workers whose jobs required frequent knee bending and at least medium levels of physical activity compared to average workers. Moreover, another study [6] showed that occupations involving kneeling for at least 30 min per day increase the radiographic knee OA associated with knee pain by 240%. Many clinical observations showed that the patellofemoral (PF) joint was a common contributor to knee pain [7], which was likely caused by excessive joint contact pressures during the high knee-flexion tasks [8, 9]. All the data from these epidemiologic

Communicated by Estefanía Peña.

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studies suggest that jobs that entail prolonged or repeated knee bending will increase the risk for knee OA [10].

Overwhelmingly, evidence indicated that overloading the knee joint causes degenerative development of the articular cartilage, leading to knee OA. Since it is difficult to directly measure the *in vivo* joint loading experimentally, musculoskeletal models have been applied to quantify muscle and joint forces in the knee, either *in vivo* [11, 12] or *in vitro* [13–15]. The knee joint loads estimated by various approaches differ substantially. During high knee-flexion postures, the thigh will come into contact with the calf, resulting in an internal contact force that was found to have a non-negligible effect [16–18]. In most of the previous models, the effects of the thigh-shank contact were not considered. Zelle et al. [16] analyzed the contact forces between the thigh and shank during high knee-flexion and found that thigh-shank contact force at around 130° of knee flexion reached 30–34% body weight (BW) at each leg. For the knees that experienced total knee arthroplasty surgery, the thigh-shank contact helped reduce the joint contact force by approximately 40% at the maximal flexion (155°) [17]. Zelle et al.'s models [16, 17] were static, considered only the tibiofemoral (TF) joint, as they did not include muscles. In a previous study [18], we analyzed the effects of the posterior thigh-shank contact on joint loading during high knee-flexion, and found that the thigh-shank contact helped reduce the PF and TF contact force by as much as 57% at a knee flexion of about 120°.

Baseball/softball catchers need to maintain deep squatting postures for an extended period of time. Knee savers have been used to alleviate knee loading during repetitive deep squatting tasks. Dooley et al. [19] investigated the effects of knee savers on the kinematics and kinetics of the lower body in deep squatting postures. The forces carried by the knee saver were measured with instrumented knee supports and the knee joint loadings were calculated via inverse dynamic modeling. Their study showed that the knee savers could carry approximately 20% BW on each side, and the use of the knee savers helped to reduce the sagittal moment in the knee by 43 and 63%, respectively, on the dominant and non-dominant side [19]. However, it is not known if the knee savers' effects in reducing joint loading in heel down squatting are as effective as in catcher squatting.

In the catcher's squat, the heel will usually be raised from the ground [20], whereas construction workers mostly squat with the heels on the ground for extended time. Previous study suggested that the joint forces in the knees are affected by the postures of the squatting or other high knee-flexion actions [21]. Lu et al. [22] analyzed the loading in the lower extremity during deep squats for different individuals performing different squatting postures. Given the study's focus on knee joint loading during occupational activities, we analyzed the deep squatting with the heels on the ground to evaluate the effects of the knee savers.

In some occupational activities, such as in floor installation, roofing, mining, gardening, and construction repairing, workers will need to perform tasks involving high knee-flexion. Different from the squat performed by a baseball or softball catcher, workers often likely keep their heels on the ground in these occupational activities [20, 22]. Knee savers that have been used in sports to reduce the knee loading during the catcher's squats have also been used in occupational activities [23]. The purpose of the current study was two-fold: first was to develop a biomechanical model that accounts for the effects of the interface contact forces between thigh and shank, or between thigh-shank and knee savers; and second was to evaluate, using the biomechanical modeling, the effects of the knee savers on the musculoskeletal loading in the knee joint in high knee-flexion tasks.

## Materials and Methods

### Experiment

The test protocol has been approved by the Institutional Review Board (IRB) of the National Institute for Occupational Safety and Health (NIOSH). Some of the experimental data have been used to evaluate the comfort of the knee savers for the high knee-flexion roofing tasks in our project [23]. The tests required the subjects to start from a standing posture, squat down to a working posture, and return to the standing posture (Fig. 1). The subjects were instructed to perform continuous, slow squats at their own comfortable pace. The subjects' heels remained on the ground during the tasks. The tasks were repeated without and with knee savers. Five healthy male subjects (age 20.6 ±1.96 years, body mass 87.74 ±9.98 kg, height 1.81 ±0.08 m) participated in the study. A pair of off-the-shelf foam knee savers (Easton, AliMed) was used in the study. The knee savers measure 20 × 11 × 11 cm and each has a mass of 0.275 kg. Kinematics for the human body during the squatting postures were determined using methods similar to that in our previous studies [23, 24]. Semi-spherical, retro-reflective markers (4 mm diameter) were applied individually on each of the body segments using thin self-adhesive tapes. A 14-camera Vicon Nexus system (Oxford Metrics Ltd., Oxford, England) was used to capture the marker trajectories at a sampling rate of 100 Hz. Although the kinematics of the whole body were measured in the tests, our focus was on the lower extremity, which consists of three segments (thigh, shank, and foot). The ground reaction forces were collected at 1200 Hz via two force plates (Model OR6-6-2000, Advanced Mechanical Technology, Inc., Watertown, MA, USA). Muscular activities of the quadriceps and hamstrings were recorded using a surface EMG system (Bagnoli, Delsys, Boston, MA, USA) for a selected participant. The EMG data were collected at a rate of 1000 Hz and synchronized with the kinematic data on the Vicon Nexus software platform.

### Determination of Thigh-Shank Contact Pressure

A knee saver contacted the human body via two contact scenarios: 1) the thigh/knee-saver and 2) shank/knee-saver contact. The contact pressures on the two surfaces of the knee saver had to be symmetric to maintain an equilibrium. Therefore, the contact pressure on only one of the contact interfaces needed to be measured. In the study [23], we selected to measure the contact pressure in the thigh/knee-saver interface, because this surface is relatively flatter and easier to place the pressure sensor film. The interface contact pressures between the thigh and shank, as well as between the thigh and knee saver, were measured using a pressure sensor film (TekScan Model 5101, 70 kPa range; Fig. 1). The pressure sensor film had a dimension of $112 \times 112$ mm and contained a $44 \times 44$ array of sensors. The resultant normal contact force ($F_n$) is obtained by the mathematical integration of contact pressure over the contact surface:

$$F_n(t) = \sum_{i=1}^{44} \sum_{j=1}^{44} p(t, x_i, y_j)\Delta x \Delta y \tag{1}$$

where $t$ and $p(t, x, y)$ are the time and pressure, respectively; and $x_i = i\Delta x$, $y_j = j\Delta y$, and $\Delta x = \Delta y = 2.5$ mm, $i,j = 1, 2, \ldots 44$.

The time-dependent center of the pressure $[x_c(t), y_c(t)]$ was calculated by:

$$\begin{aligned} x_c(t) &= \frac{1}{F_n} \sum_{i=1}^{44} \sum_{j=1}^{44} p(t, x_i, y_j) x_i \Delta x \Delta y, \text{ and} \\ y_c(t) &= \frac{1}{F_n} \sum_{i=1}^{44} \sum_{j=1}^{44} p(t, x_i, y_j) y_j \Delta x \Delta y. \end{aligned} \tag{2}$$

The contact forces between the thigh and shank or between the knee savers and the thigh-shank were considered as external forces applied onto the posterior thigh and shank in the inverse dynamic analysis. The resultant normal contact force ($F_n$) in the knee-saver-shank interface is assumed to be identical to that in the knee-saver/thigh interface. Mechanically, since the internal contact forces around the knee joints are a pair of action and reaction forces, they cancel each other (Fig. 2a). In order to maintain an equilibrium, the contact forces on the shanks and thighs should be aligned, in opposite directions, and have the same magnitude. If the resultant contact force ($F$) on one side of the body segment is decomposed into a normal ($F_n$) and a shear component ($F_f$), the orientation of $F$ relative to the the body segments is determined by $\alpha$. In any locomotion position, the angle between the thigh and shank ($\gamma$) can be equally divided by creating a bisection line (Fig. 2a). From fundamental geometry, it can be demonstrated that $\alpha = \gamma/2$. By definition, the knee-flexion angle ($\beta$) is related to $\gamma$ by $\beta = \pi - \gamma$. Consequently, the direction of $F$ can be determined uniquely based on the knee-flexion angle ($\beta$) at any locomotion position:

$$\begin{aligned} &\frac{F_f}{F_n} = \tan\alpha = \tan\frac{1}{2}\gamma = \tan\left[\frac{1}{2}(\pi - \beta)\right], \text{ and} \\ &F_f = F_n \tan\left[\frac{1}{2}(\pi - \beta)\right]. \end{aligned} \tag{3}$$

The joint forces were defined in the local coordinated systems at the joints (Fig. 2b). "2" indicates the normal contact direction within a joint, while "1" and "3" correspond to two shear directions. The orientations of the joint contact forces in the global coordinate system vary during movements; however, they consistently align with the normal contact directions within the local coordinates of the joints.

### Biomechanical Modeling

An existing, whole body human model with detailed anatomical components of the knee (AnyBody Model version 6.0, Repository 2.0) was adopted and modified. The major modification of the existing model to fit the scenario for high knee-flexion was the implementation of the contact forces between the posterior surface of the shank and thigh (Fig. 3). The knee model includes bony segments of the tibia, femur, and patella. The motions of the TF and PF joints are represented by 11-degrees-of-freedom (DOF), in which TF joint has six DOFs and PF joint has five DOFs due to the restriction of the rigid patellar ligament. The model contains the skeleton of the whole body; however, it only includes the muscles of the lower extremity. The lower extremity contains a total of 159 muscles. We are especially interested in the forces in the quadriceps and hamstrings muscles. Each of the quadriceps groups includes four muscles (rectus femoris, vastus lateralis, vastus intermedius, and vastus medialis), which were modeled with a total of 26 small muscles. Each of the hamstrings includes three muscle groups (biceps femoris, semitendinosus, and semimembranosus), which contain six small muscles. The effects of the mass and inertia properties of the whole body, which incudes the upper and lower extremity, trunk, neck, and head, were considered in the modeling. The model input consisted of the kinematics of the human body and the contact forces (i.e., $F_n$ and $F_f$) on the shanks and thighs. The muscle forces, the constraint forces in the TF and PF joints, and the ground reaction forces (GRF) were calculated using an inverse dynamic algorithm built in AnyBody system. The comparison of the predicted GRFs to those measured in the experiments served as a model validation.

### Statistical Analysis

The mean values for each of the parameters were calculated from the data of five repeats from each subject at each of the test conditions. Analysis was performed on the maximal magnitude of the parameters in each task cycle. Statistical analysis was performed with a one-tailed Student's $t$-test with a null hypothesis. Significance was set at $p < 0.05$. Furthermore, the quantitative evaluations of the effects were analyzed using the means of the parameters [25]:

$$\rho = \frac{V_1 - V_0}{V_0} \times 100\%,, \tag{4}$$

where $\rho$ is the relative variations; $V_1$ and $V_0$ imply the means of the parameters with and without knee savers, respectively. The standard deviation of $\rho$, $S_\rho$, was calculated by the Taylor approximation [25]:

$$S_\rho = \frac{V_1}{V_0}\sqrt{\left(\frac{S_0}{V_0}\right)^2 + \left(\frac{S_1}{V_1}\right)^2}, \tag{5}$$

where $S_1$ and $S_0$ are the standard deviations of the parameters $V_1$ and $V_0$, respectively.

Normality of the paired differences was assessed using the Lilliefors test and visual inspection of $Q - Q$ plots. To control for the risk of inflated Type I error across multiple outcome measures, we applied the Benjamini-Hochberg False Discovery Rate (BH FDR) procedure ($q = 0.05$) to all $p$-values. This approach is widely recommended when outcomes are correlated, as it provides a more balanced trade-off between false-positive control and statistical power than the Bonferroni correction. Effect sizes were quantified using Cohen's $d$ for paired samples, calculated as the mean of the paired differences divided by the standard deviation of those differences. This approach provides a standardized measure of effect that is directly comparable across studies, with values of 0.2, 0.5, and 0.8 conventionally interpreted as small, medium, and large, respectively.

## Results

In the experiments, subjects start from an upright posture, squat down, and then return to the upright posture. If the time period from the upright posture, when a subject starts the squatting, to the upright posture, when the subject finishes squatting, is considered as a squatting period, the entire squatting movement is normalized to a 0–100% scale. All data were normalized in the same method.

For a selected representative participant, the experimentally measured knee-flexion angles (Fig. 4a) and the normal contact forces on the thighs and shanks (Fig. 4b) as a function of the task pace without knee savers are compared with those with knee savers. For this subject, the descending action (~55% of the task pace) was approximately 22% slower than the ascending action (~45% of the task pace). The maximal knee-flexion angles (Fig. 4a) were reduced from ~ 130° for tasks without knee savers to ~ 105° with the knee savers. The peak normal contact forces on the thighs and shanks (Fig. 4b) increased from ~220 to ~350 N due to wearing the knee savers. The predicted ground reaction forces (GRF) by the model are compared with the corresponding experimental measurements for tasks without and with knee savers are shown in Fig. 4c and d, respectively. The GRFs shown were the sum of the vertical GRFs at the left and right foot, which were normalized by BW. It is seen that the model predictions agree well with the experimental data for both test conditions. The difference at the maximal knee flexion for test conditions without and with the knee savers, respectively, was approximately 0.01 and 0.03 BW.

For the same subject, the knee-flexion angle and the calculated muscle forces as a function of task pace for tasks without knee savers are compared with those with knee savers, as in Fig. 5a (knee flexion), b (quadriceps), and c (hamstrings). The patterns of the muscle force variations in a task pace look nearly symmetric in the descending and ascending periods. The peak muscle forces did not appear at the maximal knee-flexion posture, but during the squatting from the upright standing to the working posture and during the returning from the working posture to the upright standing. Wearing knee savers changed the patterns of the forces in the hamstrings muscle (Fig. 5c), however, had little effect on the patterns of the forces in the quadriceps muscle (Fig. 5b). The mean peak muscle forces in the quadriceps decreased from ~2500 to ~2000 N, whereas the peak muscle forces in the hamstrings increased marginally from ~1000 to ~1100 N, due to wearing the knee savers.

For the same subject, the knee-flexion angle, EMG signals in the quadriceps and the hamstrings are plotted as a function of the task pace in Figs. 6a, b, c, respectively. Both the quadriceps and hamstrings muscles are more active during the ascending motions than in the descending motions and at the maximal knee-flexion position (Fig. 6b, c). The EMG signals of the quadriceps and hamstrings are re-plotted as a function of knee flexion in Fig. 6d and e, respectively. The trends of the EMG signals for both squatting tasks, without and with knee savers, are consistent. There is no significant difference between the peak EMG signals for the two high knee-flexion tasks.

For the same subject, the normal contact forces, and the calculated quadriceps and hamstring muscle forces as a function of knee flexion are shown in Fig. 7a, b, c, respectively. The use of the knee savers restricted the knee-flexion motion (i.e., maximal knee-flexion angle reduced) and increased the normal contact forces on thigh and shank (Fig. 7a). The knee-flexion angles, at which the peak quadriceps forces occurred, were shifted from ~ 108° to ~ 84°, due to the use of the knee savers (Fig. 7b). However, the knee savers seem to have little effect on the pattern of the muscle forces in the hamstrings (Fig. 7c).

For the same subject, the calculated PF contact forces in the superior/inferior ($F_1$), anterior/posterior ($F_2$), and lateral/medial ($F_3$) direction are shown in Fig. 8a, c, e, respectively. The maximal PF contact force was observed in the anterior/posterior direction ($F_2$ in Fig. 8c), that is the normal contact direction between the patella and femoral segments. The predicted normalized TF contact forces in the anterior/posterior ($F_1$), superior/inferior ($F_2$), and lateral/medial ($F_3$) direction are shown in Fig. 8b, d, f, respectively. The maximal TF contact force was also in the normal contact direction in the TF joint ($F_2$ in Fig. 8d). The maximal peak contact forces in the PF joint (i.e., $F_2$, Fig. 8c) were reduced by approximately 1/3, whereas the maximal contact force in the TF joint (i.e., $F_2$, Fig. 8d) was minimally affected due to the use of the knee savers. The peak PF and TF joint contact forces (Fig. 8c, d) occurred at nearly the same knee flexion, which was reduced from ~108° to ~85° due to the use of knee savers, consistent with the trends of the quadriceps forces (Fig. 7b).

The results for all five subjects are summarized in Table 1. With $n$=5, the Lilliefors test did not reject the assumption of normality ($p > 0.05$). In addition, visual inspection of $Q-Q$ plots suggested the differences were approximately symmetric, supporting the use of the paired $t$-test. After applying the BH FDR correction, all six outcome measures remained significant (adjusted $p$-values 0.0168–0.0492), confirming that the observed differences were robust to multiple-comparison adjustment. The resultant force ($F$) and normal contact force ($F_2$) in the PF and TF joints, the normal contact forces on the thigh-shank ($F_n$), and the knee flexion angles with knee savers are significantly different from those without knee savers ($p < 0.05$). A quantitative analysis further indicated that the normal contact forces on the thigh and shanks ($F_n$)

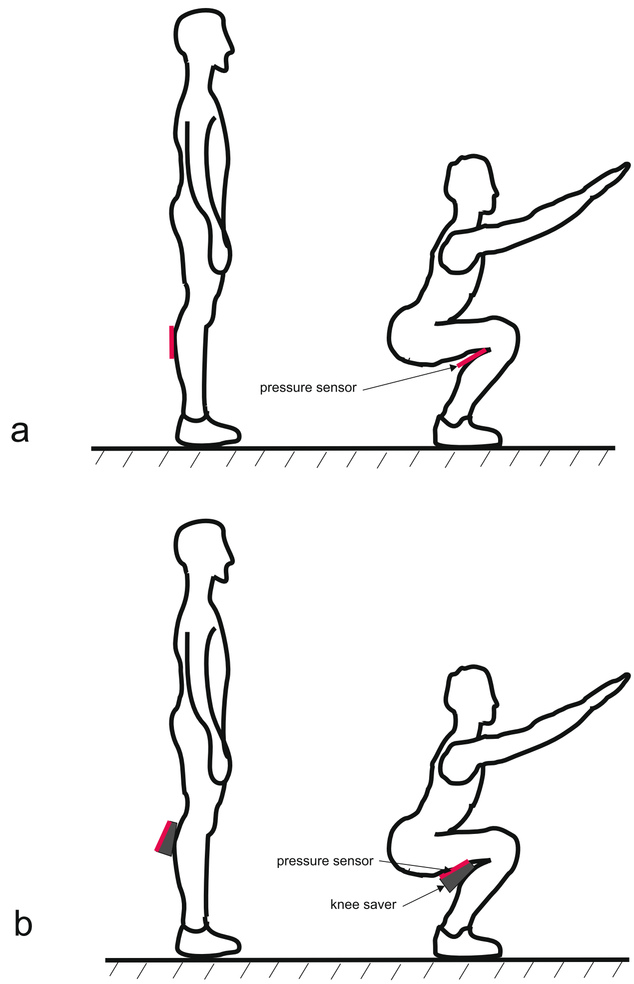


**Fig. 1** Depict of the experimental setup for basic comfort and fitting tests of knee savers. The subjects started from a standing posture, squatted down to a working posture, and returned to the standing posture. The subjects' heels remained on the ground during the actions. The actions were repeated without (**a**) and with knee savers (**b**). The pressure was measured in the thigh-shank contact interface (**a**) and in the thigh/knee-saver contact interface (**b**) using a pressure sensor film

**Table 1** Summary of the results of five participants

| Parameters | w/o KS (mean) | (STD) | w KS (mean) | (STD) | $p$-value | $\rho$ (%) | $S_\rho$ (%) |
|---|---|---|---|---|---|---|---|
| Knee Flexion (deg) | 104.7 | 11.4 | 98.3 | 8.6 | 0.0492 | −6.1 | 1.7 |
| PF Joint $F$ (N) | 3059.9 | 376.6 | 2597.5 | 358.0 | 0.0099 | −15.1 | 2.3 |
| PF Cont $F_2$ (N) | 2635.0 | 313.2 | 2104.3 | 315.2 | 0.0118 | −20.1 | 2.1 |
| TF Joint $F$ (N) | 2514.5 | 427.5 | 2367.4 | 476.7 | 0.0097 | −5.9 | 6.2 |
| TF Cont $F_2$ (N) | 2106.9 | 309.5 | 1988.7 | 327.6 | 0.0028 | −5.6 | 4.3 |
| Thigh-shank Cont $F_n$ (N) | 190.2 | 59.7 | 271.2 | 62.5 | 0.0097 | 42.6 | 36.8 |

The resultant forces in PF and TF joints are defined as $F=\sqrt{F_1^2+F_2^2+F_3^2}$. The normal contact forces in the PF and TF joints are in the superior/inferior direction, $F_2$, as defined in Fig. 2b. The relative variations of the parameters ($\rho$) were calculated via Eqs. (4) and (5). "w/o KS" and "w KS" represent test conditions without and with knee savers, respectively. Significance was set at $p < 0.05$

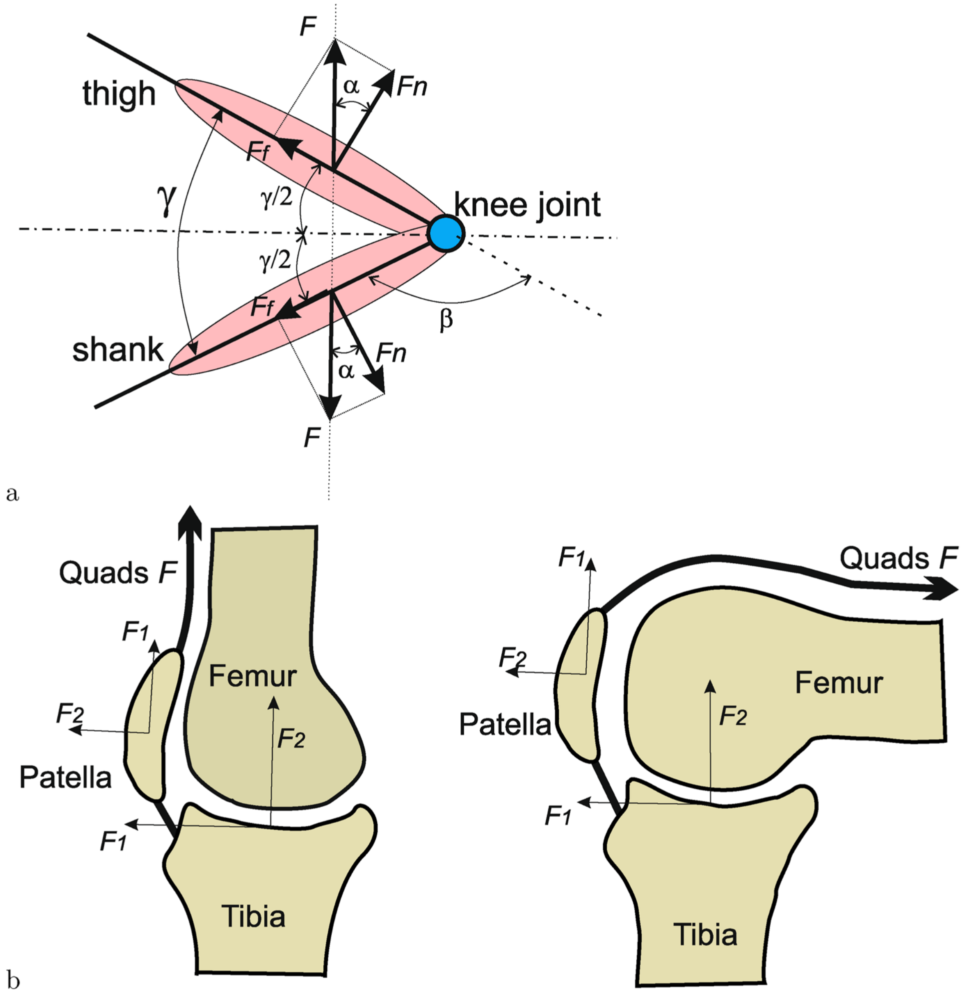


**Fig. 2** Determination of the contact forces in the thigh-shank and thigh/knee-saver interfaces and the biomechanical modeling. **a** Determination of the contact forces in the thigh-shank and thigh-knee-saver interfaces. The contact force ($F$) on the thigh or shank is decomposed of a normal ($F_n$) and a shear component ($F_f$). **b** The definitions of the force directions in PF and TF joints. "1", "2", and "3" represent the local coordinates defined at the joints. "2" indicates the normal contact direction (superior/inferior) within a joint, while "1" and "3" correspond to two shear directions (i.e., anterior/posterior and lateral/medial). A joint force is decomposed into $F_1$, $F_2$, and $F_3$ in the locale coordinate systems. The normal joint contact forces were in direction "2", represented by $F_2$; and the resultant shear forces were $\sqrt{F_1^2 + F_3^2}$

increased by ~42.6%, and the resultant force and normal contact force in the PF joint were reduced by ~15.1 and ~ 20.1%, respectively, due to the knee savers. The use of the knee savers would slightly reduce the TF joint loading (i.e., $F$ and $F_2$) by nearly 6%. Cohen's $d$ values for the force outcomes ranged from −2.42 to +1.69, indicating very large to extremely large effects across these measures. Wearing knee savers helped reduce the knee flexion angle by ~6.1%, however, the results were barely significant ($p = 0.0492$).

## Discussion

Specific occupational activities, such as roofing, floor installation, and mining, require workers to maintain high knee-flexion postures for extended time. These awkward working postures have been associated with the knee OA among workers [3–6, 10]. Some workers wear knee savers to reduce the knee joint loading during occupational activities that require squatting and kneeling. Most workers feel more comfortable at their knees during high knee-flexion tasks when they use knee savers [23]. In order to improve the product design, we analyzed the biomechanics of the knee savers in relation to musculoskeletal loading. Our model predictions showed that the peak muscle forces and joint contact forces did not appear at the maximal knee-flexion posture, but occurred during the squatting from the upright standing to the working posture and during the returning from the working posture to the upright standing. The current results indicated that wearing knee savers during high knee-flexion tasks helped to reduce the peak contact forces in the PF joint by nearly 20%. Since it is generally known that excessive contact pressures in the PF joint may cause the degeneration of the joint [8, 9] and that the ailments in the PF joint were a common cause for knee pains in clinical observations [7], wearing knee savers may potentially help reduce the risks of the knee OA for workers who are frequently required to perform high knee-flexion tasks for extended time.

In our study, the normal contact forces on the thigh and shank were measured to be approximately 0.22 and 0.33 BW, respectively, for high knee-flexion tasks without and with knee savers. These results are consistent with a previous

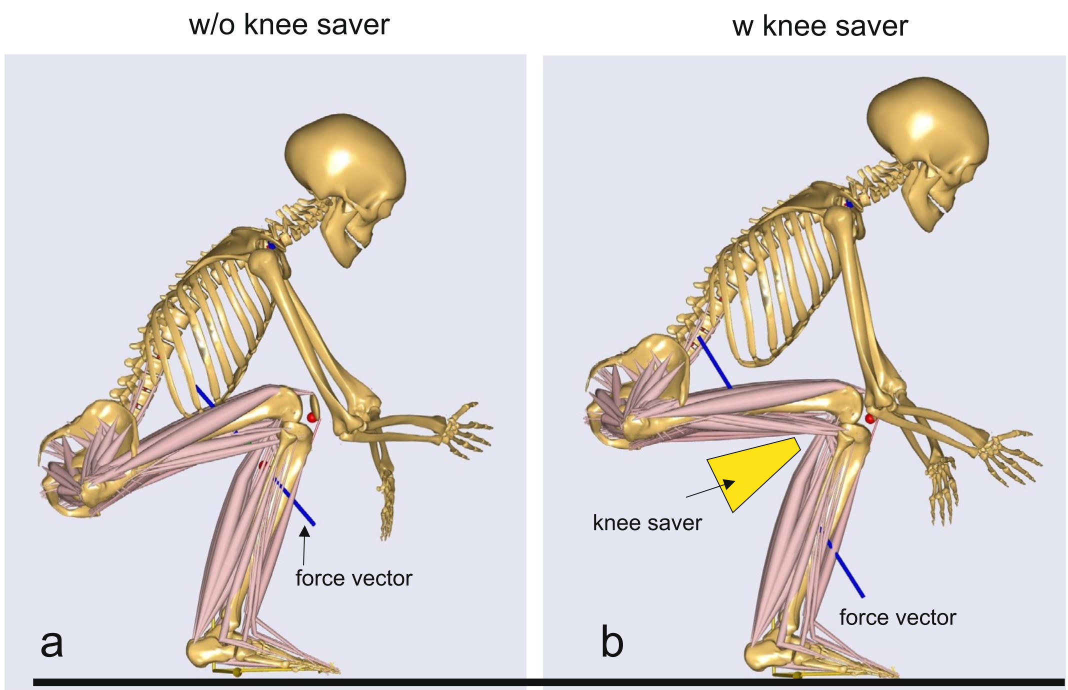


**Fig. 3** Biomechanical model used in the study. **a** High knee-flexion posture without knee savers. **b** High knee-flexion posture with knee savers. The human model was adopted from AnyBody (version 7.2). The skeletons of the whole body were included in the modeling; however, only the lower extremities of the model included muscles

study [16], but higher than that reported by Dooley et al. [19], who found the forces on knee supports during catcher squatting tasks to be about 0.18 BW. These variations are likely due to the differences in the measurement methods and squatting postures. In Dooley et al.'s [19] study, the loads carried by the knee savers were not measured directly, but estimated by an instrumented knee supporter. The instrumented knee supporter was placed between the thigh and shank, in a location close to the ankle. In their measurement setup, the subject's thigh was still in contact with the shank during the squatting task, i.e., only a portion of the thigh-shank contact forces was carried by the instrumented knee supporter, which was measured in their study. In addition, in the current study the subjects' heels remained on the ground (close to the Asian squatting posture), whereas the heels were raised above the ground during the catcher squatting posture used in the previous study [19]. Previous studies [20, 22] indicated that Asian squatting postures required higher knee flexion compared to catcher's squatting. Higher knee-flexion angles will cause larger contact forces between the thighs and shanks.

One of the technical challenges in the evaluation of the knee joint during high knee-flexion tasks is the experimental quantification of the contact force between the thigh and shank, or between the thigh-shank and knee savers, and the inclusion of these contact forces into the modeling. The interface contact pressures, with or without knee savers, were measured using pressure sensor films in our study. However, the pressure sensor films can only measure the normal pressure and cannot measure the shear force in the contact interface. In our previous study [18], this problem was resolved by assuming that contact forces applied on the shanks and on the thighs were aligned and in opposite directions. The shear forces in the contact interfaces were determined by a fit and try method in our previous study [18]. In the current study, we improved the method and derived the dependence of the shear force component in the contact interface on the knee-flexion angle analytically. If the knee-flexion angles and the normal contact forces are known, the shear force components in the contact interfaces can be determined. This improvement in the modeling resulted in the improved efficiency and reliability of the numerical calculations.

The wearing of the knee savers helped reduce the knee joint loading likely via two aspects. First, the wearing of the knee savers increased the pressures in the contact interfaces on the thigh and shanks. The increased contact forces on the thigh and shanks help reduce the joint loadings. Second, the wearing of the knee savers restricted the joint flexion angles. The reduction of the knee joint flexion angles helped reduce the muscle forces and joint loading.

The test results showed that reduction of the knee-flexion angle due to the knee saver was barely statistically significant ($p$ is close to 0.05, Table 1). Given the small sample size, this result should be interpreted cautiously. By a close examination of the raw data, we found that, of the five

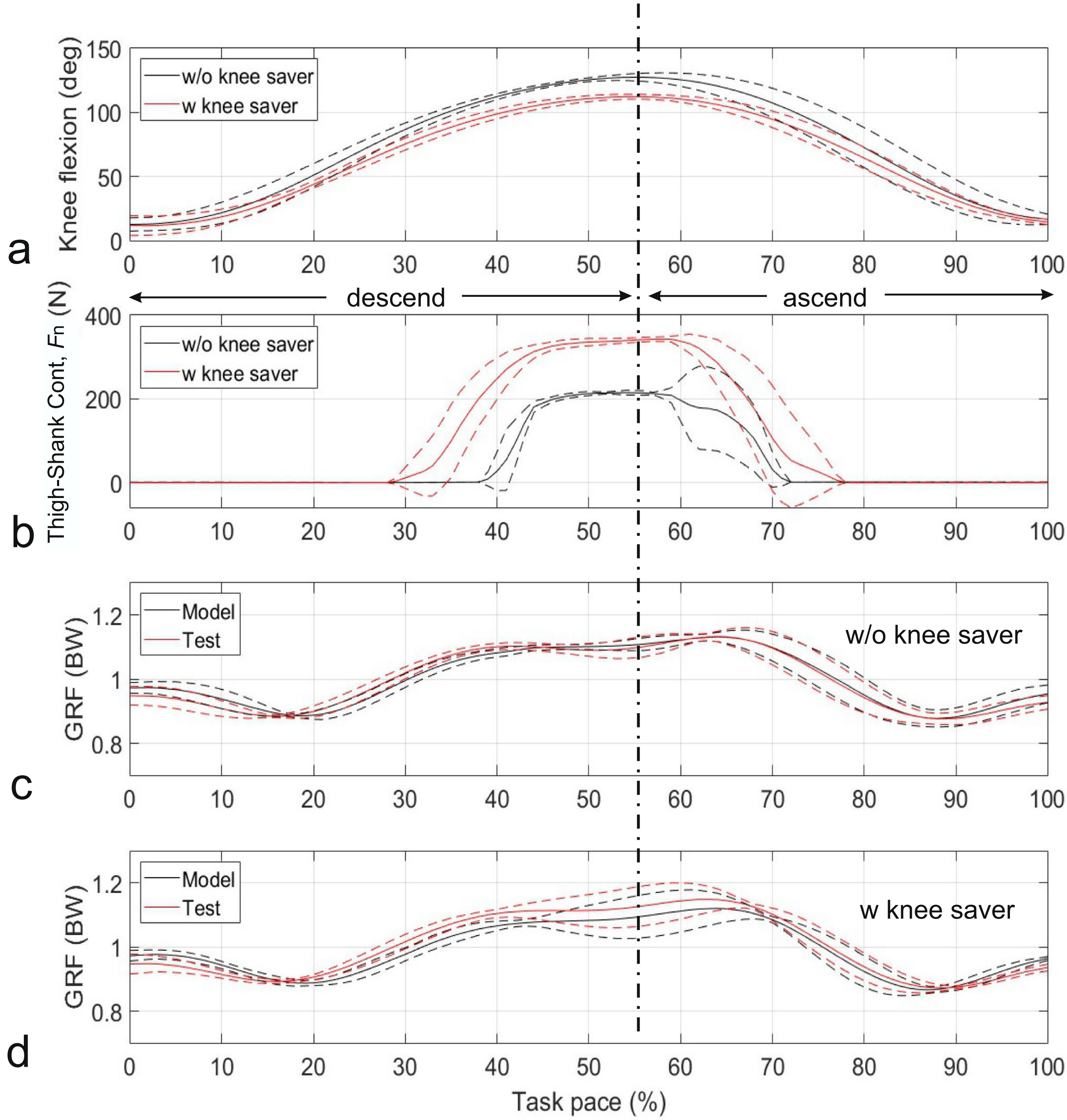


**Fig. 4** Representative experimental measurements and comparison of the simulated ground reaction forces with experiments. **a** Experimentally measured knee-flexion angles as a function of task pace without (black lines) and with knee savers (red lines). **b** Experimentally measured normal contact forces on the thigh and shank as a function of task pace without (black lines) and with knee savers (red lines). **c** Comparison of the ground reaction forces (GRF) calculated using the model ("Model", black lines) with those measured experimentally ("Test", red lines) for high knee-flexion tasks without knee savers. **d** Comparison of the ground reaction forces (GRF) calculated using the model ("Model", black lines) with those measured experimentally ("Test", red lines) for high knee-flexion tasks with knee savers. The GRFs shown were the sum of the vertical GRFs at the left and right foot, which were normalized by BW. In all plots, the solid and dashed lines represent the mean and standard deviations of the parameters, respectively. The descending phase implies the period from the standing posture to the high knee-flexion working posture. The ascending phase implies the period from the high knee-flexion working posture returning to the standing posture

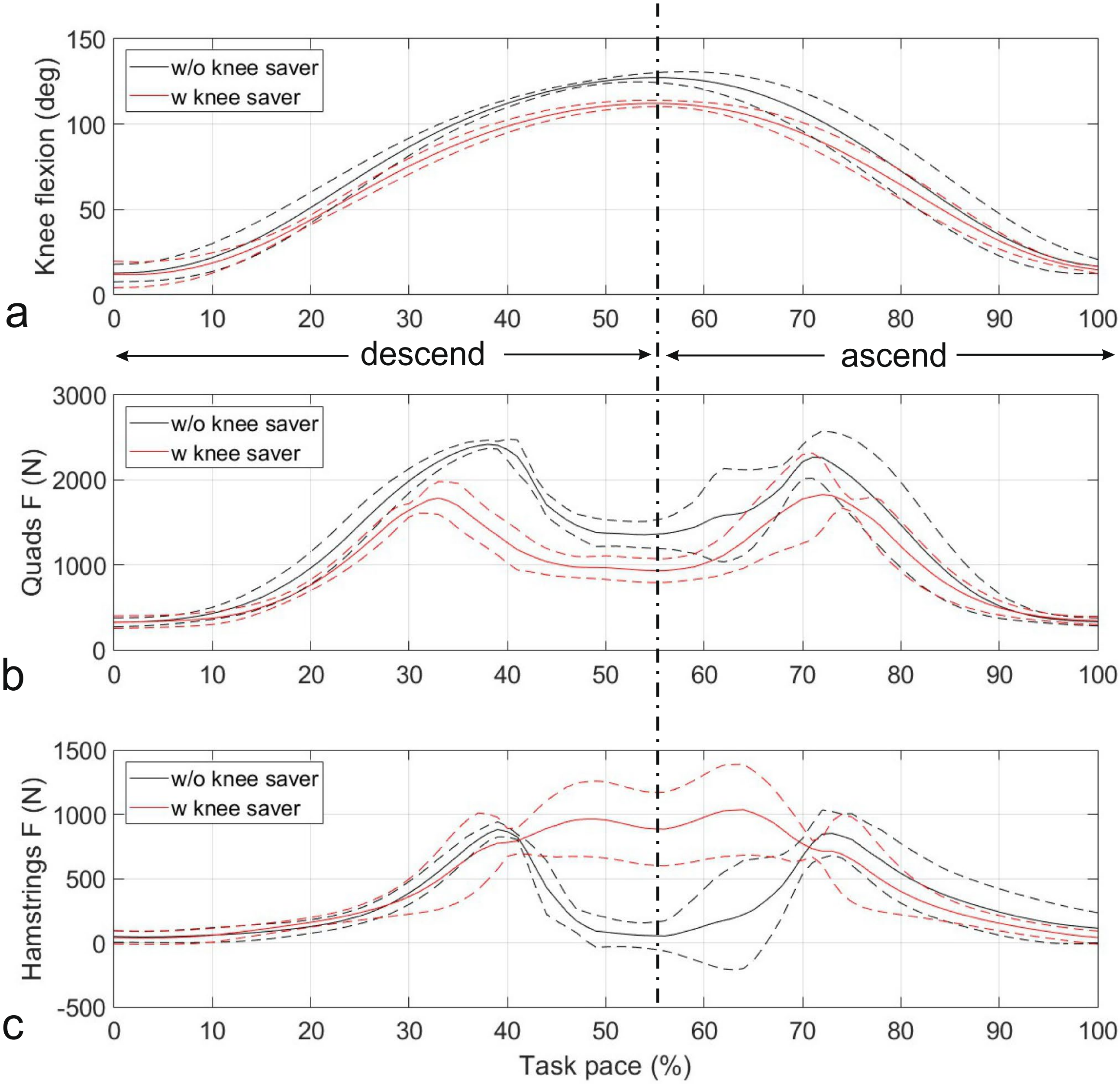


**Fig. 5** Representative results of experimentally measured knee flexion and model predictions on the muscle forces in the quadriceps and hamstrings. **a** Comparison of the experimentally measured knee-flexion angles as a function of task pace without knee savers (black lines) with those with knee savers (red lines). **b** Calculated muscle forces in the hamstrings as a function of task pace without (black lines) and with knee savers (red lines). **c** Calculated muscle forces in the quadriceps as a function of task pace without (black lines) and with knee savers (red lines). In all plots, the solid and dashed lines represent the mean and standard deviations of the parameters, respectively. The descending phase implies the period from the standing posture to the high knee-flexion working posture. The ascending phase implies the period from the high knee-flexion working posture to the standing posture

subjects, wearing knee savers helped reduce the knee-flexion angle in four subjects and slightly increased the knee-flexion angle in one subject. This variability may reflect individual movement strategies, such as not squatting to the deepest positions without knee savers. Nevertheless, the overall pattern is consistent with the expected mechanical constraint imposed by the knee savers, which physically limit the range of knee joint motion. Importantly, the corresponding effect size for knee-flexion angle was large (Cohen's $d = -0.96$), suggesting that the observed reduction is meaningful despite the limited statistical power.

The comparison of the GRFs predicted using the models with the corresponding experimental measurements served as a qualitative validation in our study. At a static upright

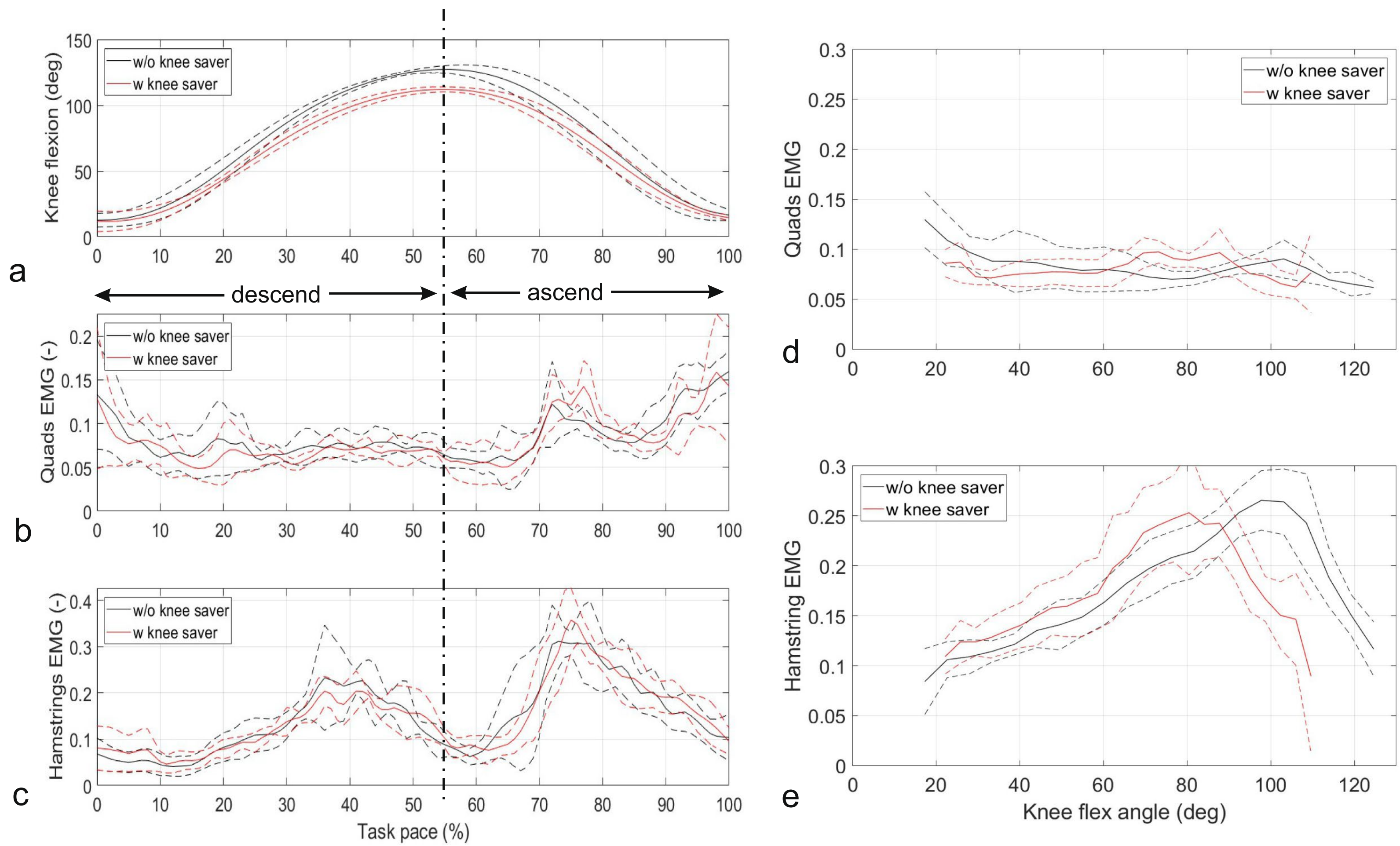


**Fig. 6** Representative EMG signals of the quadriceps and hamstrings muscles as a function of task pace and knee-flexion angles. **a** Comparison of the experimentally measured knee-flexion angles as a function of task pace without knee savers (black lines) with those with knee savers (red lines). **b** Comparison of the experimentally measured EMG of the quadriceps muscles as a function of task pace without knee savers (black lines) with those with knee savers (red lines). **c** Comparison of the experimentally measured EMG of the hamstrings muscles as a function of task pace without knee savers (black lines) with those with knee savers (red lines). **d** Comparison of the experimentally measured EMG of the quadriceps muscles as a function of knee-flexion angle without knee savers (black lines) with those with knee savers (red lines). **e** Comparison of the experimentally measured EMG of the hamstrings muscles as a function of knee-flexion angle without knee savers (black lines) with those with knee savers (red lines). In plots (**a–c**), the descending phase implies the period from the standing posture to the high knee-flexion working posture; and the ascending phase implies the period from the high knee-flexion working posture to the standing posture. In all plots, the solid and dashed lines represent the mean and standard deviations of the parameters, respectively

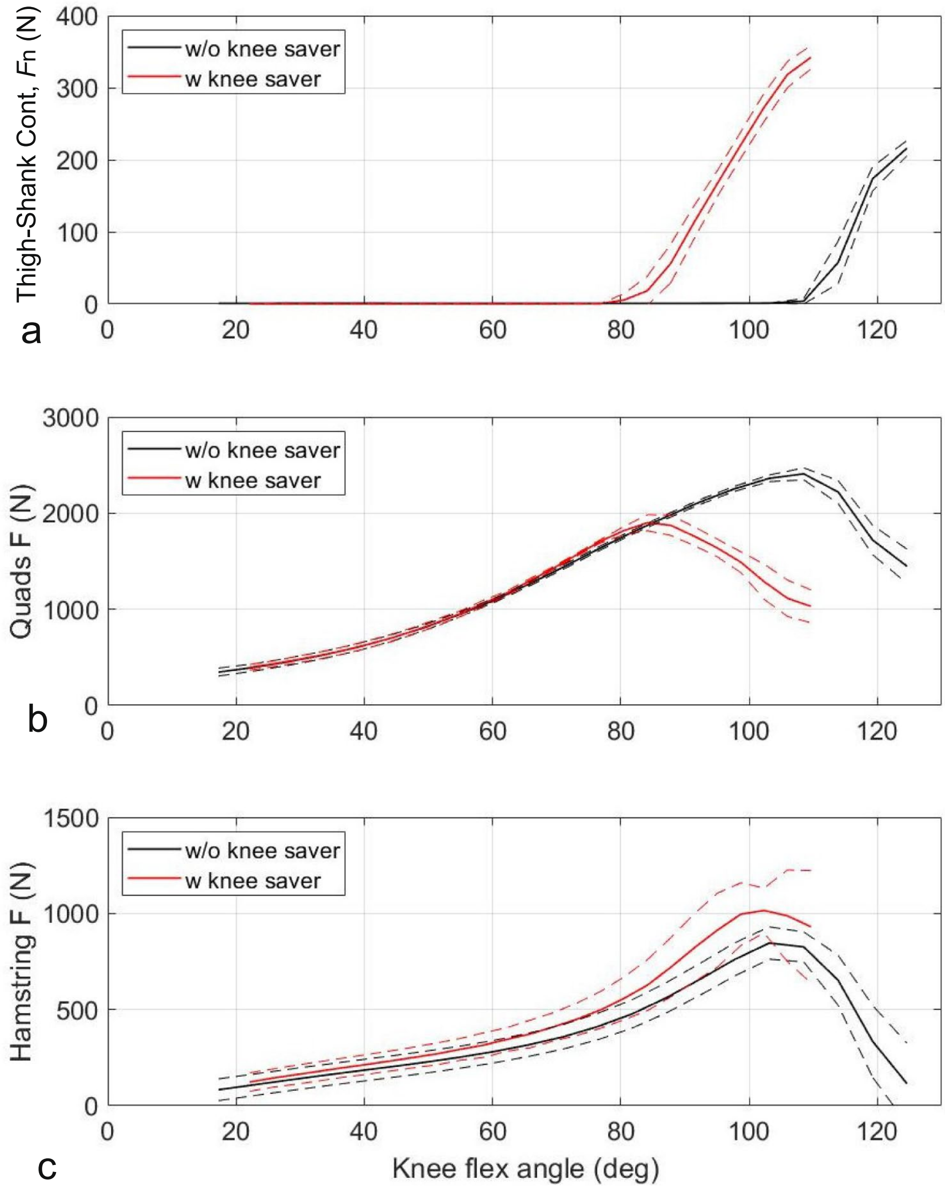


**Fig. 7** Representative results of the experimentally measured contact forces on thigh and shank and the calculated forces in the quadriceps and hamstrings muscles as a function of knee-flexion angle. **a** Comparison of the experimentally measured normal contact forces on the thigh and shank as a function of knee-flexion angle without knee savers (black lines) with those with knee savers (red lines). **b** Comparison of the calculated quadriceps muscle force as a function of knee-flexion angle without knee savers (black lines) with that with knee savers (red lines). **c** Comparison of the calculated hamstrings muscle force as a function of knee-flexion angle without knee savers (black lines) with that with knee savers (red lines). In all plots, the solid and dashed lines represent the mean and standard deviations of the parameters, respectively

standing posture, the GRF must be 1.0 BW, which was used for the model calibration (results not shown). The reason that both measured and calculated GRFs in Fig. 4c, d did not reach 1.0 BW at paces 0 and 100% is because the subject was in motion and was not in a static upright standing posture. The good agreement between the predicted GRFs and the experimental measurements indicated that the biomechanical modeling of the human body is basically reasonable and reliable. The fluctuation of the ground reaction force during the squatting task is due to the mass inertial effects of the body segments during the motions, as well as the effects of the muscular forces that are needed to maintain the dynamic balance.

The general patterns of the EMG signals for the quadriceps and hamstrings during the tasks (Fig. 6) are consistent with the model predictions on the forces in the quadriceps and hamstrings (Fig. 5). The EMG signals during a task pace (Fig. 6b, c) indicated more muscle activities for the quadriceps and hamstrings during the descending and ascending motions than at the deepest knee-flexion posture. These observations are consistent with the model predictions on the muscle forces (Fig. 5) that the quadriceps and hamstrings applied more forces during the descending and ascending motions than at the deepest knee-flexion posture. However, the EMG measurements failed to provide quantitative validations for the model predictions. The model predictions indicated that the peak quadriceps muscle forces without knee savers were higher than those with knee savers, whereas there were no noticeable differences in EMG signals collected for the tasks with and without knee savers

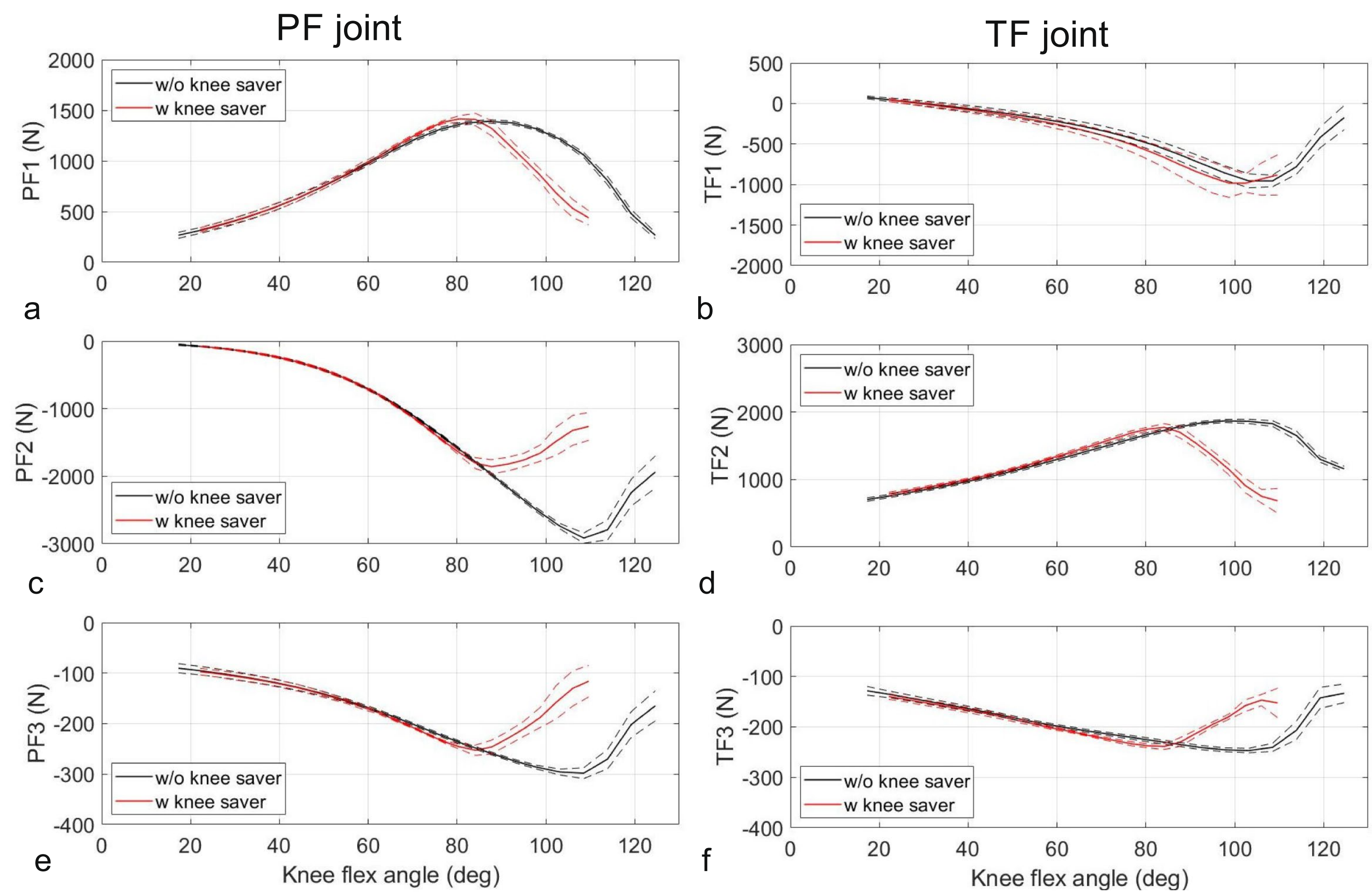


**Fig. 8** Representative results of the calculated PF and TF joint forces as a function of knee-flexion angles. **a**, **c**, **e** Comparison of the calculated PF joint forces in three dimensions (i.e., *PF*1, *PF*2, and *PF*3) as a function of knee-flexion angle without knee savers (black lines) with those with knee savers (red lines). **b**, **d**, **f** Comparison of the calculated TF joint forces in three dimensions (i.e., *TF*1, *TF*2, and *TF*3) as a function of knee-flexion angle without knee savers (black lines) with those with knee savers (red lines). The definition of the orientation of the components of PF and TF joint forces are illustrated in Fig. 2b. In all plots, the solid and dashed lines represent the mean and standard deviations of the parameters, respectively

(Fig. 6). These results are not surprising since the relationships between the EMG signals and muscles forces are also dependent on the muscles' contractile conditions [26]. For a given EMG signal, the muscle's force outputs for the eccentric, concentric, or isometric contractions are different. When the knee joints move close to the end of their ranges of motions (e.g., close to the deepest knee-flexion positions for tasks without knee savers), the muscles' contractile statuses are about to change, the variations of the muscle forces may not be proportional to the EMG signals.

In the tests, the subjects were requested to perform slow, continuous deep squats, at their own natural paces. If subjects were asked to perform the squatting tasks at a specific pace, they may be forced to squat in unnatural ways, introducing undesirable effects. In the experiments, the average time period for a squat for all participants was recorded within a range from 1.8 to 2.2 s. In general, movement speed will have effects on muscle forces. However, for the current study, the effects of the mass inertia of body segments and the speed-dependencies of the muscles on the musculoskeletal forces were minimal for these slow speed and small acceleration levels. Consequently, the impact of the inter-subject pace variations on our results should be considered negligible.

In our experiments, the internal/external rotations of the knee joints cannot be determined precisely, however, the flexion/extension and abduction/adduction motions of the knee joints were precisely determined. In kinematics, the knee savers' constraining effects on the inter/external rotation of the knee joints are minimal. Consequently, experimental measurements and the associated inverse dynamical analysis conducted in this study have successfully captured the overall trends regarding the effects of knee savers on the knee musculoskeletal loading.

A further limitation of the study is that the number of subjects and the age/gender demographics chosen do not accurately represent the general population. Nevertheless, for the purpose of biomechanical analysis, we expected to identify general trends concerning the influence of knee savers on musculoskeletal loading in the knees, necessitating only a minimal amount of experimental data as model parameters. Although the absolute values of musculoskeletal loading obtained from the study may vary among the general population, the trends identified regarding the impact of knee savers on joint loading should be widely relevant.

In summary, we investigated the effects of the knee savers on the musculoskeletal loading of the knee joint during high knee-flexion tasks using an inverse dynamic analysis. Our analysis indicated that wearing knee savers in high knee-flexion tasks would reduce the quadriceps force and the PF joint contact forces in the knee by about 20%. Our results indicated that wearing knee savers may potentially help reduce the risks of the knee OA for workers who are frequently required to perform high knee-flexion tasks.

**Acknowledgements** Authors supported by occupational safety research funds.

**Author Contributions** LZ involved in the experimental development and execution, the model development, and manuscript preparation. JZW contributed to the study design, model development, and human subject testing. KDM, ALH, REC, and CMW were engaged in human subject testing and data analysis. SPB was responsible for developing the human subject testing protocol, conducting the tests, and processing the data. All co-authors played an active role in the preparation of the manuscript.

**Funding** The project was supported by the National Institute for Occupational Safety and Health (NIOSH) through an internal funding source (# 3939kk4).

**Data Availability** A data repository associated with the manuscript is available.

## Declarations

**Conflict of interest** The authors declare no conflict of interest.

**Disclaimers** The findings and conclusions in this report are those of the authors and do not necessarily represent the official position of the National Institute for Occupational Safety and Health, Centers for Disease Control and Prevention of the USA. Mention of brand name does not constitute product endorsement.

**Ethical approval** The human subject testing protocol was approved by the Institutional Review Board (IRB) of the National Institute for Occupational Safety and Health (NIOSH). Subjects read and completed informed consents. The study was performed in accordance with the ethical standards as laid down in the 1964 Declaration of Helsinki and its later amendments or comparable ethical standards.

## References

1. Zhang, Y., and J. M. Jordan. Epidemiology of osteoarthritis. *Clin Geriatr Med*. 26:355–69, 2010.
2. CDC. A national public health agenda for osteoarthritis,. update. Centers for Disease Control and Prevention: Technical report, p. 2010, 2020.
3. Coggon, D., P. Croft, S. Kellingray, D. Barrett, M. McLaren, and C. Cooper. Occupational physical activities and osteoarthritis of the knee. *Arthritis Rheum*. 43:1443–9, 2000.
4. Anderson, J. J., and D. T. Felson. Factors associated with osteoarthritis of the knee in the first national health and nutrition examination survey (Hanes I). evidence for an association with overweight, race, and physical demands of work. *Am J Epidemiol*. 128:179–89, 1988.
5. Felson, D. T., M. T. Hannan, A. Naimark, J. Berkeley, G. Gordon, P. W. Wilson, and J. Anderson. Occupational physical demands, knee bending, and knee osteoarthritis: results from the Framingham study. *J Rheumatol*. 18:1587–92, 1991.
6. Cooper, C., T. McAlindon, D. Coggon, P. Egger, and P. Dieppe. Occupational activity and osteoarthritis of the knee. *Ann Rheum Dis*. 53:90–3, 1994.

7. Osborne, J. D., S. B. Luczak, W. B. Acker, and J. Bicos. Patellofemoral joint contact pressures: Current concepts and use in patellar instability studies. *Orthopedics*. 42:e172–e179, 2019.
8. Dutton, R. A., M. J. Khadavi, and M. Fredericson. Patellofemoral pain. *Phys Med Rehabil Clin N Am*. 27:31–52, 2016.
9. Grelsamer, R. P., and J. R. Klein. The biomechanics of the patellofemoral joint. *J Orthop Sports Phys Ther*. 28:286–98, 1998.
10. Wang, X., T. A. Perry, N. Arden, L. Chen, C. M. Parsons, C. Cooper, L. Gates, and D. J. Hunter. Occupational risk in knee osteoarthritis: A systematic review and meta-analysis of observational studies. *Arthritis Care Res (Hoboken)*. 72:1213–1223, 2020.
11. Singerman, R., J. Berilla, M. Archdeacon, and A. Peyser. In vitro forces in the normal and cruciate-deficient knee during simulated squatting motion. *J Biomech Eng*. 121:234–42, 1999.
12. Mason, J. J., F. Leszko, T. Johnson, and R. D. Komistek. Patellofemoral joint forces. *J Biomech*. 41:2337–48, 2008.
13. Dahlkvist, N. J., P. Mayo, and B. B. Seedhom. Forces during squatting and rising from a deep squat. *Eng Med*. 11:69–76, 1982.
14. Sharma, A., F. Leszko, R. D. Komistek, G. R. Scuderi, J. Cates, H. E. and F. Liu. In vivo patellofemoral forces in high flexion total knee arthroplasty. *J Biomech*. 41:642–8, 2008.
15. Komistek, R. D., T. R. Kane, M. Mahfouz, J. A. Ochoa, and D. A. Dennis. Knee mechanics: a review of past and present techniques to determine in vivo loads. *J Biomech*. 38:215–28, 2005.
16. Zelle, J., M. Barink, R. Loeffen, M. De Waal Malefijt, and N. Verdonschot. Thigh-calf contact force measurements in deep knee flexion. *Clin Biomech (Bristol, Avon)*. 22:821–6, 2007.
17. Zelle, J., M. Barink, M. De Waal Malefijt, and N. Verdonschot. Thigh-calf contact: does it affect the loading of the knee in the high-flexion range? *J Biomech*. 42:587–93, 2009.
18. Wu, J. Z., C. S. Pan, and B. M. Wimer. Evaluation of the shock absorption performance of construction helmets under repeated top impacts. *Engineering Failure Analysis*. 96:330–339, 2019.
19. Dooley, E., J. Carr, E. Carson, and S. Russell. The effects of knee support on the sagittal lower-body joint kinematics and kinetics of deep squats. *J Biomech*. 82:164–170, 2019.
20. Fox, J. and M. Hefzy. Knee and ankle biomechanics during squatting with heels on and off of the ground, with and without weight shifting. Bone and Muscle pp. 1–9, 2019.
21. Cohen, Z. A., H. Roglic, R. P. Grelsamer, J. H. Henry, W. N. Levine, V. C. Mow, and G. A. Ateshian. Patellofemoral stresses during open and closed kinetic chain exercises. *An analysis using computer simulation. Am J Sports Med*. 29(4):480–7, 2001.
22. Lu, Y., Q. Mei, H.-T. Peng, J. Li, C. Wei, and Y. Gu. A comparative study on loadings of the lower extremity during deep squat in asian and caucasian individuals via opensim musculoskeletal modelling. BioMed Research International , 2020.
23. Breloff, S. P., A. Dutta, E. W. Sinsel, R. E. Carey, C. M. Warren, F. Dai, S. Ning, and J. Z. Wu. Are knee savers and knee pads a viable intervention to reduce lower extremity musculoskeletal disorder risk in residential roofers? *Int J Ind Ergon*. 74: 2019.
24. Breloff, S. P., A. Dutta, F. Dai, E. W. Sinsel, C. M. Warren, X. Ning, and J. Z. Wu. Assessing work-related risk factors for musculoskeletal knee disorders in construction roofing tasks. *Appl Ergon*.81:102901, 2019.
25. Wolter, K. M. Taylor Series Methods. Introduction to Variance Estimation, New York: Springer, pp. 221–247, 1985.
26. Hof, A. Emg and muscle force: an introduction. *Human Movement Science*. 3:119–153, 1984.