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Effects of second generation and indoor sports surfaces on knee joint kinetics and kinematics during 45° and 180° cutting manoeuvres; and exploration using statistical parametric mapping and Bayesian analyses.

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Effects of second generation and indoor sports surfaces on knee joint kinetics and kinematics during 45° and 180° cutting manoeuvres; and exploration using statistical parametric mapping and Bayesian analyses.

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Abstract

Purpose: The aim of the current investigation was to examine the influence of second generation (2G) and indoor surfaces on knee joint kinetics, kinematics, frictional and muscle force parameters during 45° and 180° change of direction movements using statistical parametric mapping (SPM) and Bayesian analyses.

Methods: Twenty male participants performed 45° and 180° change of direction movements on 2G and indoor surfaces. Lower limb kinematics were collected using an eight-camera motion capture system and ground reaction forces were quantified using an embedded force platform. ACL, patellar tendon and patellofemoral loading was examined via a musculoskeletal modelling approaches and the frictional properties of the surfaces were examined using ground reaction force information. Differences between surfaces were examined using SPM and Bayesian analyses.

Results: Both SPM and Bayesian analyses showed that ACL loading parameters were greater in the 2G condition in relation to the indoor surface. Conversely, SPM and Bayesian analyses confirmed that patellofemoral/ patellar tendon loading alongside the coefficient of friction and peak rotational moment were larger in the indoor condition compared to the 2G surface.

Conclusions: This study indicates that the indoor surface may improve change of direction performance owing to enhanced friction at the shoe-surface interface but augment the risk from patellar tendon/ patellofemoral injuries; whereas the 2G condition may enhance the risk from ACL pathologies.

Introduction

The benefits of physical activity/ sport are unequivocal [1] and initiatives to improve participation are commonplace [2]. However, sports/ physical activity is associated with a high incidence of injuries [3, 4]. The annual cost of treating sports injuries in high school athletes alone is estimated to be >$2 billion [4], with 1/5 school children absent at least one day per year [6] and 1/3 working adults losing at least one day a year to sports injuries [6].

Importantly, Hootman et al., [7] showed in their aetiological examination of 15 different sports, that the lower extremities were the most common location for injury. Specifically, the knee joint is the most commonly injured musculoskeletal structure in athletes, accounting for over 30% of all reported sports injuries [8]. The most frequently reported knee condition in sports medicine clinics is patellofemoral pain which has a prevalence cited between 22.7-28.9% [9], and manifests as dull retropatellar pain, aggravated by activities that frequently and excessively load the joint [10]. Chronic patellar tendinopathy is also a common musculoskeletal condition that presents clinically as localised pain at the proximal tendon attachment [11]. Aetiological analyses have shown that the incidence of patellar tendinopathy may be as high as 36%, with this specific condition accounting for as many as 25% of all soft tissue injuries [12]. Tendinopathy is mediated through excessively forces at the patellar tendon itself, with failed reparative response due to insufficient rest between loading exposures [11]. Similarly, the anterior cruciate ligament (ACL) is the most frequently reported acute sports injury [13], with over 175,000 ACL reconstruction surgeries being performed each year in the US alone [14]. ACL injuries are predominantly non-contact in nature, whereby the ligament becomes compromised in the absence of physical contact between athletes [15]. Mechanically, ACL injuries occur when the ligament experiences excessive tensile forces [16].

Given the prevalence and debilitating nature of sports injuries, considerable research attention has been devoted to modifying the underlying mechanisms linked to the aetiology of common sports-related pathologies. It has been strongly advocated that the properties of sports playing surfaces can influence both the performance of athletes and the likelihood of injury occurrence [17]. Traditionally most sports were played on natural surfaces, however, owing to climatic and economic factors, artificial alternatives have become increasing popular over the past 30 years, with synthetic grass and indoor surfaces being the most commonly encountered [18]. Many athletic disciplines involve sprinting, stopping and rapid changes in movement direction [19]. Frictional torque generated at the shoe-surface interface means that the knee may be subjected to excessive stresses when rapid directional changes are undertaken [20]. Therefore, the level of traction between the shoe and surface is one of the most commonly cited factors influencing lower limb injury occurrence [21].

There are concerns that some of the mechanical properties of artificial sports-surfaces may be associated with acute and chronic knee injuries in relation to natural surface, and there is evidence from descriptive epidemiological analyses to support this notion [22]. However, clinical, biomechanical and epidemiological analyses have focused heavily on the differences between performing on natural vs. synthetic surfaces, and there has yet to be a published investigation examining different synthetic surfaces on the biomechanical mechanisms linked specifically to the aetiology of knee pathologies.

Furthermore, whilst aetiological literature has through prospective and retrospective analyses been able to identify the risk factors linked to the aetiology of knee pathologies, these biomechanical parameters are explored in the scientific literature through discrete point analysis [23]. For time normalized biomechanical parameters, statistical parametric mapping (SPM) may represent a more efficacious process, as it is able to examine an entire time-based data sequence and reduces the likelihood of a type II error by eliminating the need for multiple analyses [23]. Similarly, in recent years Bayesian assessments have gained considerable acceptance and practicability, although in spite of their prospective benefits [24], their utilization in biomechanical analyses remains limited. To date there has yet to be a biomechanical investigation examining the effects of different synthetic surfaces on the biomechanical parameters linked to the aetiology of knee pathologies using a concurrent SPM and Bayesian approach.

Therefore, the aim of the current investigation was to examine the influence of second generation (2G) and indoor surfaces on knee joint kinetics, kinematics, frictional and muscle force parameters during 45° and 180° change of direction movements using SPM and Bayesian analyses.

Methods

Participants

Twenty male recreational athletes (age = 23.00±2.51years, stature = 176.22±8.36cm and mass = 76.79±10.60kg) volunteered to take part in this study. The procedure utilized for this investigation was approved by an institutional ethical committee. All participants were free from musculoskeletal pathology at the time of data collection and had not previously undergone knee surgery. Written informed consent was obtained in accordance with the principles outlined in the Declaration of Helsinki.

Surfaces

The data collection took place in an indoor biomechanics laboratory. The indoor surface (MondoSport Ramflex, Mondo, Italy) had a total thickness of 6 mm, with a vulcanized rubber construction. The indoor surface was comprised of a 2 mm surface layer and a 4 mm base layer and was mounted over an underlying concrete surface. The 2G surface utilized for this investigation was an 8 mm polyethylene, synthetic turf. For the 2G surface condition, the turf was strong affixed to the existing laboratory surface and force platform using double-sided carpet tape. Following the completion of their data collection protocol, participants were asked to subjectively indicate which surface that they preferred, and which surface they felt provided more traction.

Procedure

Participants completed five repeats of two sport-specific movements 45° and 180° change of direction in the two surface conditions. To control for any order effects, the order in which participants performed in each movement and surface condition was counterbalanced. Kinematic information was obtained using an eight-camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) using a capture frequency of 250 Hz. Dynamic calibration of the system was performed before each data collection session. Calibrations producing residuals <0.85 mm and points above 4000 in all cameras were considered acceptable. To measure kinetic information an embedded piezoelectric force platform (Kistler National Instruments, Model 9281CA) operating at 1000 Hz was utilised. The kinetic and kinematic information were synchronously obtained using an analogue to digital board and interfaced using Qualisys track manager.

Lower extremity segments were modelled in 6 degrees of freedom using the calibrated anatomical systems technique [25]. To define the segment co-ordinate axes of the right foot, shank and thigh, retroreflective markers were placed unilaterally onto the 1st metatarsal, 5th metatarsal, calcaneus, medial and lateral malleoli, medial and lateral epicondyles of the femur. To define the pelvis segment further markers were positioned onto the anterior (ASIS) and posterior (PSIS) superior iliac spines. Carbon fiber tracking clusters were positioned onto the shank and thigh segments (Figure 1). The foot was tracked using the 1st metatarsal, 5th metatarsal and calcaneus markers and the pelvis using the ASIS and PSIS markers. The centers of the ankle and knee joints were delineated as the mid-point between the malleoli and femoral epicondyle markers, whereas the hip joint centre was obtained using the positions of the ASIS markers. Static calibration trials (not normalized to static trial posture) were obtained in each footwear allowing for the anatomical markers to be referenced in relation to the tracking markers/ clusters. The Z (transverse) axis was oriented vertically from the distal segment end to the proximal segment end. The Y (coronal) axis was oriented in the segment from posterior to anterior. Finally, the X (sagittal) axis orientation was determined using the right-hand rule and was oriented from medial to lateral (Figure 2).

@@@FIGURE 1 NEAR HERE@@@

@@@FIGURE 2 NEAR HERE@@@

Data were collected during the 45° and 180° change of direction movements as described below:

45° change of direction

Participants completed 45° change of direction movements using an approach velocity of 4.0 m/s ± 5% striking the force platform with their right (dominant) limb. Cut angles were measured from the centre of the force plate and the corresponding line of movement was delineated using masking tape so that it was clearly evident to participants. The stance phase of this movement was defined as the duration over > 20 N of vertical ground reaction force (GRF) was applied to the force platform.

180° change of direction

Participants completed 180° change of direction movements using an approach velocity of 4.0 m/s ± 5% striking the force platform with their right (dominant) limb, then returning in the initial direction of travel. The stance phase of this movement was defined as the duration over > 20 N of vertical GRF was applied to the force platform.

Processing

Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). All data were linearly normalized to 100% of the stance phase. GRF and kinematic data were smoothed using cut-off frequencies of 50 and 12 Hz with a low-pass Butterworth 4th order zero lag filter [26]. Three-dimensional kinematics of the knee were calculated using an XYZ cardan sequence of rotations (where X=sagittal plane; Y=coronal plane and Z=transverse anatomical planes). Joint moments were computed using Newton–Euler inverse-dynamics, allowing net knee joint moments to be calculated. To quantify joint moments segment mass, segment length, GRF and angular kinematics were utilized.

Patellofemoral loading was quantified using a model adapted from van Eijden et al., [27], in accordance with the protocol of Willson et al., [28] in that co-contraction of the knee flexor musculature was accounted for. This musculoskeletal model has been shown to be sufficiently sensitive to resolve differences between different footwear [29], across different foot orthotic configurations [30], different prophylactic knee sleeves [31], between sexes [32, 33] and between those with and without patellofemoral pain [34]. Hamstring and gastrocnemius forces were calculated in accordance with previously established procedures [35]. Hamstring and gastrocnemius forces (N) were multiplied by their moment arms relative to the knee flexion angle [36], and then summed to generate a knee flexor moment. The knee flexor moment was added to the net knee extensor moment quantified using inverse dynamics and divided by the quadriceps moment arm [27], to obtain quadriceps force (N) adjusted for co-contraction of the knee flexors. From the above processing quadriceps and hamstring force (N·s) impulses during the stance phase were extracted using a trapezoidal function. Quadriceps and hamstring force (N/s) load rates were also extracted by obtaining the peak increase in force between adjacent data points.

Patellofemoral force (N) was quantified in accordance with the protocol of van Eijden et al., [27]. Patellofemoral joint stress (MPa) was quantified by dividing the patellofemoral force by the patellofemoral contact area. Patellofemoral contact areas were obtained in accordance with the sex specific data of Besier et al., [37]. From the above processing patellofemoral force (N·s) and stress (MPa·s) impulses during the stance phase were extracted using a trapezoidal function. Patellofemoral force (N/s) and stress (MPa/s) load rates were also extracted by obtaining the peak increase in force/ stress between adjacent data points using the first derivative function in Visual 3D.

In addition, patellar tendon loading was quantified using a musculoskeletal model similarly adapted from Janssen et al., [38]. This model has been shown to be sufficiently sensitive to resolve differences in patellar tendon kinetics between different prophylactic knee sleeves [39], between sexes [40], different rehabilitation mechanisms [41] and between dominant and non-dominant limbs [42]. Once again, the derived knee flexor moment was added to the net knee extensor moment quantified using inverse dynamics, and then divided by the moment arm of the patellar tendon, generating the patellar tendon force. The sex specific tendon moment arms were quantified using the data of Herzog & Read, [43]. From the above processing, patellar tendon force (N·s) impulse during the stance phase was extracted using a trapezoidal function. Patellar tendon load rate (N/s) and was also extracted by obtaining the peak increase in force between adjacent data points using the first derivative function in Visual 3D.

ACL loading was similarly quantified using a musculoskeletal modelling approach as described and validated by Dai and Yu, [44]. This approach has been shown to be sufficiently sensitive to resolve differences in ACL force during different movements [44], different prophylactic knee sleeves [45], between sexes [46] and also as a function of different athletic footwear [20]. The face validity of this current model has been evaluated from two key aspects in the scientific literature. Firstly, Dai and Yu, [44] showed that the model exhibited a high level of consistency with values provided from in vivo ACL loading [47]. Secondly, the timing of ACL rupture in dynamic tasks occurs ≤50 ms following initial foot contact [48]. The timing of the peak ACL force estimated using this model by Dai and Yu, [44] and Sinclair and Taylor [45] shown to be <50 ms, is therefore consistent with this data and further supports the face validity of the model. From the above processing, ACL force (N·s) impulse during the stance phase was extracted using a trapezoidal function. ACL load rate (N/s) and was also extracted by obtaining the peak increase in force between adjacent data points. Further, to the above the knee abduction moment impulse (Nm·s) during the stance phase was extracted using a trapezoidal function and the abduction moment load rate (Nm/s) was also extracted by obtaining the peak increase between adjacent data points using the first derivative function in Visual 3D.

Finally, the loading rates (N/s) of the vertical and braking GRFs were also extracted by obtaining the peak increase in vertical and anterior-posterior GRF between adjacent data points. Furthermore, the peak translation coefficient of friction (μ) of each footwear was determined from the ratio of horizontal and vertical force components during the initial period of shoe motion [20]. The peak rotational moment of the GRF (Nm) was used to describe the rotational friction characteristics of the footwear [49].

Following this, three-dimensional knee joint kinematics, vertical GRF, anterior posterior GRF, quadriceps force, hamstring force, patellofemoral force, patellofemoral stress, patellar tendon, ACL and knee abduction moment parameters were extracted during the entire stance phase and time normalized to 101 data points using linear interpolation for each participant.

Statistical analyses

Differences across the entire stance phase were examined using 1-dimensional statistical parametric mapping (SPM) with MATLAB 2018a (MATLAB, MathWorks, Natick, USA), in accordance with Pataky et al., [50], using the source code available at http://www.spm1d.org/. Differences between surfaces were examined using paired t-tests (SPM t). The alpha (α) level for statistical significance for SPM was set at the 0.05 level.

Differences in discrete biomechanical parameters that could not be contrasted using SPM were examined using Bayesian factors (BF) to explore the extent to which the data supported the alternative (H_1) or null (H_0) hypotheses i.e. that there were or were no meaningful differences between surface conditions for both males and females. Bayes factors were interpreted in accordance with the recommendations of Jeffreys, [51], with values <1 indicating no evidence, 1-3 anecdotal evidence, 3-10 indicating substantial evidence, 10-30 strong evidence, 30-100 very strong evidence and >100 decisive evidence in support of H_1. In accordance with the aforementioned recommendations, values >3 were considered sufficient evidence in support of H_1. Finally, participants' subjective ratings were examined using Chi-squared (X^2) tests. Discrete statistical tests were conducted using SPSS v25.0 (SPSS, USA).

Results

Knee joint kinetics, kinematics, muscle forces and GRFs contrasted using SPM are presented in figures 3-4 and the discrete parameters are found in tables 1-2.

@@@TABLE 1 NEAR HERE@@@

@@@TABLE 2 NEAR HERE@@@

45° change of direction

Statistical parametric mapping

ACL force was shown to be significantly greater in the 2G surface from 15-25 and 70-100 % of the stance phase (Figure 3D). The SPM analyses also showed that patellar tendon force, patellofemoral force, patellofemoral stress and quadriceps force was significantly greater in the indoor surface from 15-75 % of the stance phase (Figure 3E-H). In addition, hamstring forces (Figure 3I) were significantly greater in the 2G surface from 0-50 and 85-95 % of the stance phase and braking forces were significantly larger in the indoor surface from 0-5 and 20-80 % of the stance phase (Figure 3K). Finally, the knee abduction moment was shown to be significantly greater in the 2G surface from 10-95 % of the stance phase (Figure 3L).

@@@FIGURE 3 NEAR HERE@@@

Discrete parameters

For the translational coefficient of friction, quadriceps integral and peak rotational moment there was decisive evidence in favour of these parameters being greater in the indoor condition. There was also strong evidence in favour of the quadriceps force load rate being greater in the indoor condition. Furthermore, for the hamstring integral and hamstring load rate there was decisive evidence in favour of these parameters being greater in the 2G surface (Table 1).

Furthermore, for the patellar integral, patellofemoral force integral and patellofemoral stress integral there was decisive evidence in favour of these parameters being greater in the indoor condition. There was also substantial-strong evidence that the patellar load rate, patellofemoral force load rate and patellofemoral stress load rates were larger in the indoor surface. Finally, for the ACL integral and knee abduction moment integral there was decisive evidence in favour of these parameters being greater in the 2G condition and strong evidence in favour of the knee abduction moment load rate being larger in the 2G surface (Table 2).

180° change of direction

Statistical parametric mapping

ACL force was shown to be significantly greater in the 2G surface from 5-95 % of the stance phase (Figure 4D). The SPM analyses also showed that patellar tendon force, patellofemoral force, patellofemoral stress and quadriceps force was significantly greater in the indoor surface from 20-30 % of the stance phase (Figure 4E-H). In addition, hamstring forces (Figure 4I) were significantly greater in the 2G surface from 15-20 and 70-95 % of the stance phase and braking forces were significantly larger indoor surface from 10-90 % of the stance phase (Figure 4K). Finally, the knee abduction moment was shown to be significantly greater in the 2G surface from 10-90 % of the stance phase (Figure 4L).

@@@FIGURE 4 NEAR HERE@@@

Discrete parameters

For the translational coefficient of friction, load rate braking and peak rotational moment there was decisive evidence in favour of these parameters being greater in the indoor condition. Furthermore, for the hamstring integral and there was decisive evidence in favour of these parameters being greater in the 2G surface (Table 1).

Finally, for the ACL integral and knee abduction moment integral there was decisive evidence in favour of these parameters being greater in the 2G condition and strong evidence in favour of the knee abduction moment load rate being larger in the 2G surface (Table 2).

Subjective ratings

For the subjectively preferred surface, the chi-squared test was non-significant (X^2 = 0.20, P>0.05) with nine participants reporting a preference for the indoor surface and eleven for the 2G surface. However, for the subjective ratings of surface traction the chi-squared test was significant (X^2 = 5.00, P<0.05), with fifteen participants reporting that the indoor surface provided more traction and five participants indicating the 2G surface.

Discussion

The aim of the current investigation was to examine the effects of 2G and indoor surfaces on knee joint kinetics, kinematics, frictional and muscle force parameters during 45° and 180° change of direction movements using SPM and Bayesian analyses. To the authors knowledge this is the first investigation of this nature and may provide further important information regarding the effects of different surfaces on the risk factors linked to the aetiology of knee pathologies during functional athletic tasks.

This investigation importantly confirmed that the coefficient of friction and peak rotational moment were greater in the indoor surface in relation to the 2G condition. This observation is in agreement with the subjective ratings, which similarly showed that participants rated that the indoor surface provided more traction. As there were notable differences in braking force parameters observed using SPM and Bayesian analyses yet no differences in vertical GRFs, it can be concluded that alterations in the coefficient of friction were mediated through alterations in anterior-posterior GRFs. Importantly, frictional forces allow the resultant GRF vector to be directed more effectively towards the intended direction, mediating enhanced linear acceleration [52]. However, increased traction has also been linked to the aetiology of injury [20, 53]. Nonetheless, owing to an enhanced coefficient of friction this observation suggests that the indoor surface mediated an increased resistance to sliding compared to the 2G condition.

Importantly the current investigation showed using SPM and Bayesian analyses that ACL loading parameters were greater in the 2G condition, an interesting observation as hamstring forces were larger and quadriceps forces were reduced in the 2G condition. As the quadriceps load the ACL by mediating anterior tibial translation, and the hamstrings oppose tibial translation and thus act to offload the ACL [20, 54], it could be expected that ACL forces would be attenuated in the 2G surface. However, recent subject- specific musculoskeletal modelling investigations have shown that the knee abduction moment is the inverse-dynamic mechanism that most strongly governs the magnitude of ACL loading [55]. Indeed, the knee abduction moment influences ACL loading by altering the tolerance of the ACL to anterior tibial translation forces [55] and has been shown through a prospective in vivo investigation as the biomechanical factor that most strongly predicts ACL injury [56]. It can therefore be conjectured that the enhanced ACL loads experienced in the 2G surface conditions were mediated via the enhanced knee abduction moments that were also revealed in this condition. Importantly, the aetiology of ACL injuries in athletic populations is linked to excessive loading of the ACL itself [16]. Therefore, owing to an enhanced ACL loading in the 2G surface, the findings indicate that the specific 2G surface examined in this investigation may increase the risk from ACL injury during sport specific change of direction movements compared to the indoor surface.

In addition, the current study also showed using SPM and Bayesian analyses that both patellofemoral and patellar tendon loading were larger in the indoor condition compared to the 2G surface. It is proposed that these observations were mediated through the increased quadriceps forces in the 2G surface, as previous analyses have shown that quadriceps kinetics strongly affect patellar tendon/ patellofemoral loading [27, 38]. This observation concurs with the conclusions of Yu et al., [57] who showed that an enhanced coefficient of friction directly increases the force of contraction from the quadriceps. This observation may be clinically important as excessive patellar tendon/ patellofemoral joint loading are the mechanisms most strongly linked to the aetiology of pain symptoms in active individuals [10, 11]. It can be concluded on account of the enhanced tendon/ joint loading that the indoor surface examined in the current investigation may increase the risk from chronic knee pathologies injury during change of direction movements.

A potential limitation that should be acknowledged of the current investigation is that only male athletes were examined. Female athletes have been shown to exhibit distinct external joint moments [58], ACL loading [26, 46], lower extremity joint kinematics [58] and patellofemoral joint stress [32] compared to male athletes. This suggests that further investigation into the effects of different surfaces using a female sample is warranted before comprehensive conclusions can be drawn.

In conclusion, although previous investigations have examined the biomechanical influence of different surfaces, current knowledge regarding the effects of 2G and indoor surfaces on the is biomechanics of change of direction movements is limited. As such the current investigation contributes to biomechanical literature by providing a comprehensive examination of knee joint kinetics, kinematics, frictional and muscle force parameters during 45° and 180° change of direction movements. Importantly, this study showed using both SPM and Bayesian analyses that ACL loading parameters were greater in the 2G condition in relation to the indoor surface. Conversely, SPM and Bayesian analyses confirmed that patellofemoral/ patellar tendon loading alongside the coefficient of friction and peak rotational moment were larger in the indoor condition compared to the 2G surface. This study indicates that the indoor surface may improve change of direction performance owing to enhanced friction at the shoe-surface interface but augment the risk from patellar tendon/ patellofemoral injuries whereas the 2G condition may enhance the risk from ACL pathologies.

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Figure labels

Figure 1: Experimental marker configuration.

Figure 2: Pelvic, thigh, tibial and foot segments, with segment co-ordinate system axes (X=sagittal plane; Y=coronal plane and Z=transverse anatomical planes).

Figure 3: Kinetic and kinematic parameters as a function of surface for the 45° change of direction movement (Black = 2G & Red = Indoor).

Figure 4: Kinetic and kinematic parameters as a function of surface for the 180° change of direction movement (Black = 2G & Red = Indoor). (Black = 2G & Red = Indoor).

Table 1: Frictional and muscle force parameters (mean±SD) as a function of the experimental movement and surface conditions.

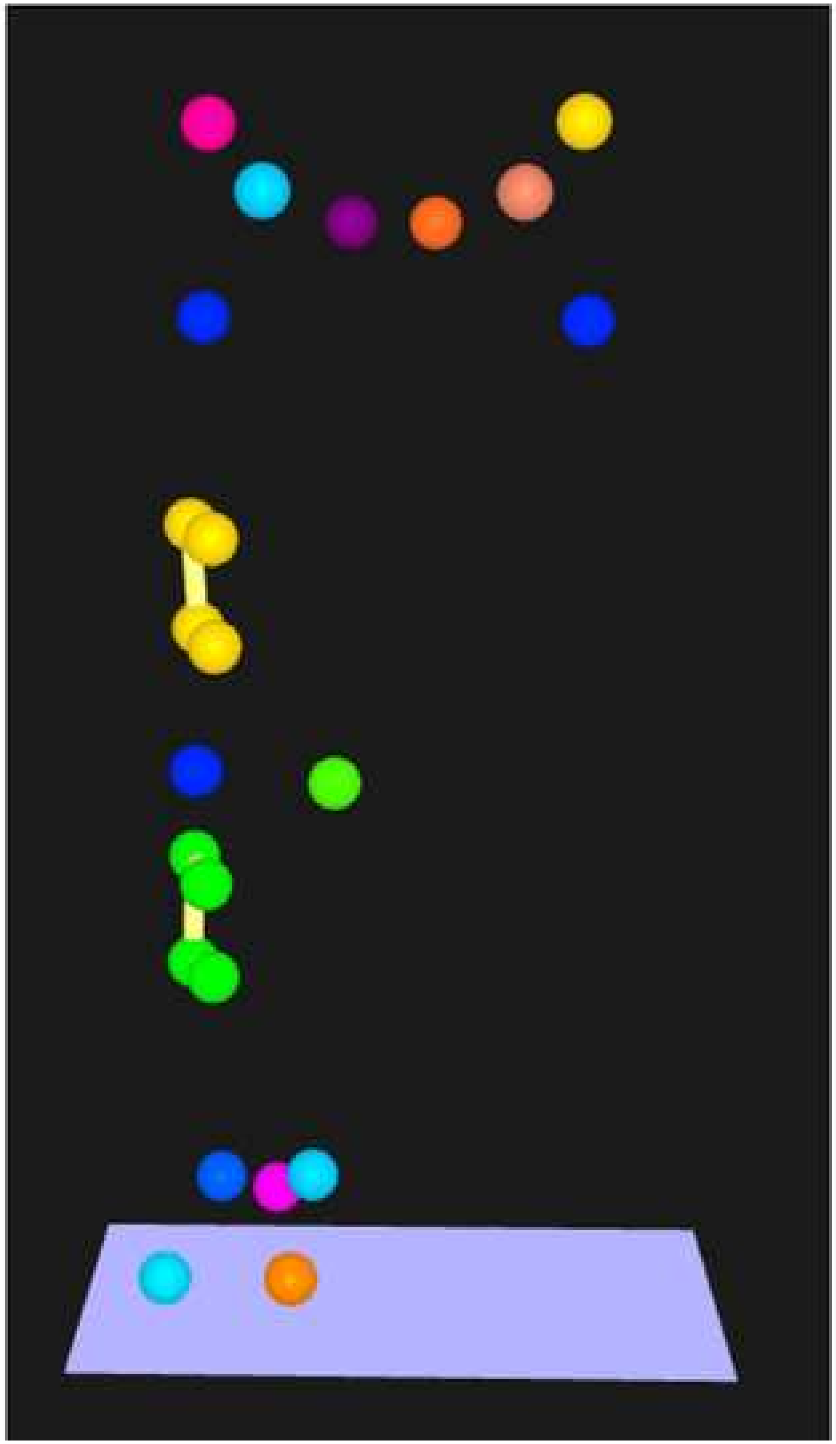
	45°				Bayes factor	180°				Bayes factor
	2G		**Indoor**			**2G**		**Indoor**		
	Mean	*SD*	**Mean**	*SD*		**Mean**	*SD*	**Mean**	*SD*	
Translational coefficient of friction (μ)	0.18	0.06	0.43	0.22	***1955.58***	0.31	0.04	0.68	0.12	***3790913148***
Load rate vertical GRF (N/s)	257380.63	145386.71	213585.60	166827.07	*0.98*	91427.29	44069.39	90583.25	41206.03	*0.22*
Load rate braking (N/s)	105688.64	68411.50	113758.53	88612.97	*0.25*	41308.94	25208.46	57044.95	25020.64	***256.22***
Quadriceps integral (N·s)	519.07	205.05	799.63	327.81	***2122.58***	1232.12	617.79	1356.81	441.27	*0.37*
Quadriceps force load rate (N/s)	294927.37	89591.22	400281.59	184320.64	***13.81***	182912.39	69116.06	215434.26	73472.04	*1.45*
Hamstring integral (N·s)	418.28	245.20	172.63	73.43	***700.97***	776.07	357.91	485.60	299.91	***114.61***
Hamstring load rate (N/s)	451553.79	262740.43	355276.42	218654.34	***231.17***	163808.19	68777.20	160825.39	62659.50	*0.22*
Peak rotational moment (Nm)	10.71	4.95	17.23	7.73	***843.31***	6.35	2.23	19.86	7.66	***405689***

Notes: Bold Bayes factors indicate values >3, i.e. at least substantial evidence in support of H_1.

Table 2: Knee joint loading parameters (mean±SD) as a function of the experimental movement and surface conditions.

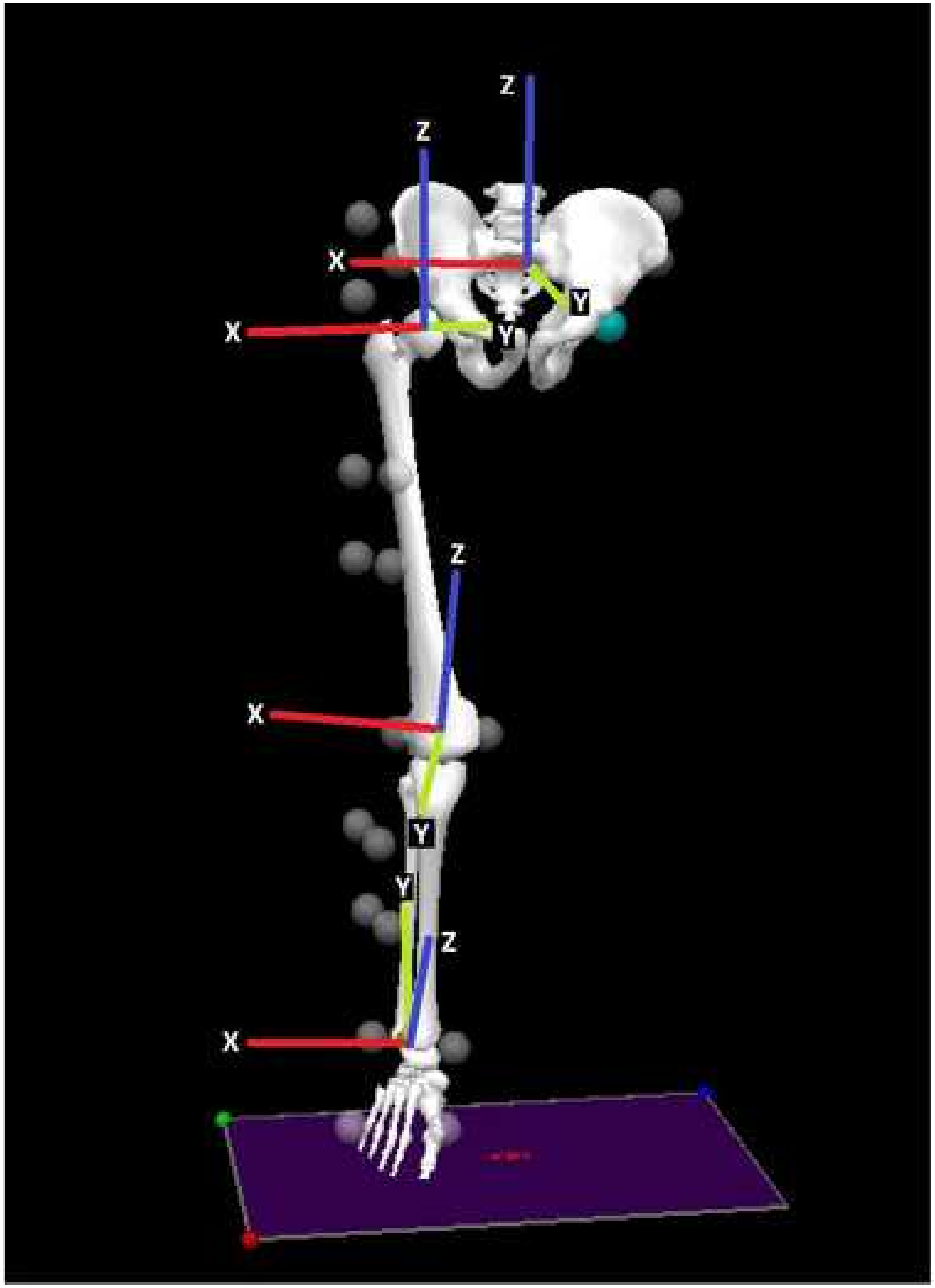
	45°				***Bayes factor***	180°				***Bayes factor***
	2G		**Indoor**			**2G**		**Indoor**		
	Mean	*SD*	**Mean**	*SD*		**Mean**	*SD*	**Mean**	*SD*	
ACL integral (N·s)	120.04	60.17	60.12	29.30	***184.85***	569.31	157.10	336.58	103.41	***9738740***
ACL load rate (N/s)	130835.62	98683.33	101851.46	105484.49	*2.92*	151352.59	79027.39	136652.04	85639.02	*0.39*
Patellar integral (N·s)	368.81	126.16	556.94	201.50	***2656.68***	808.91	390.88	897.91	288.06	*0.43*
Patellar load rate (N/s)	245518.44	75613.85	346615.91	189618.36	***5.31***	146763.38	51995.62	181376.58	77201.91	*1.63*
Patellofemoral force integral (N·s)	272.08	112.17	419.84	178.95	***1545.97***	661.33	336.80	724.64	239.71	*0.35*
Patellofemoral force load rate (N/s)	149456.58	46041.49	201387.17	90435.87	***16.12***	93317.93	35693.47	109493.77	36830.67	*1.49*
Patellofemoral stress integral (MPa·s)	0.47	0.16	0.72	0.27	***3993.67***	1.04	0.51	1.16	0.37	*0.45*
Patellofemoral stress load rate (MPa/s)	295.30	88.25	409.72	207.64	***7.63***	179.01	64.70	216.02	81.14	*1.62*
Knee abduction moment integral (Nm·s)	14.28	8.50	-3.65	3.34	***4436895***	73.01	27.13	13.71	13.44	***81161120***
Knee abduction moment load rate (Nm/s)	13815.90	7780.47	9684.01	7381.42	***28.85***	15426.31	7695.95	12834.24	10020.31	*0.55*

Notes: Bold Bayes factors indicate values >3, i.e. at least substantial evidence in support of H_1.

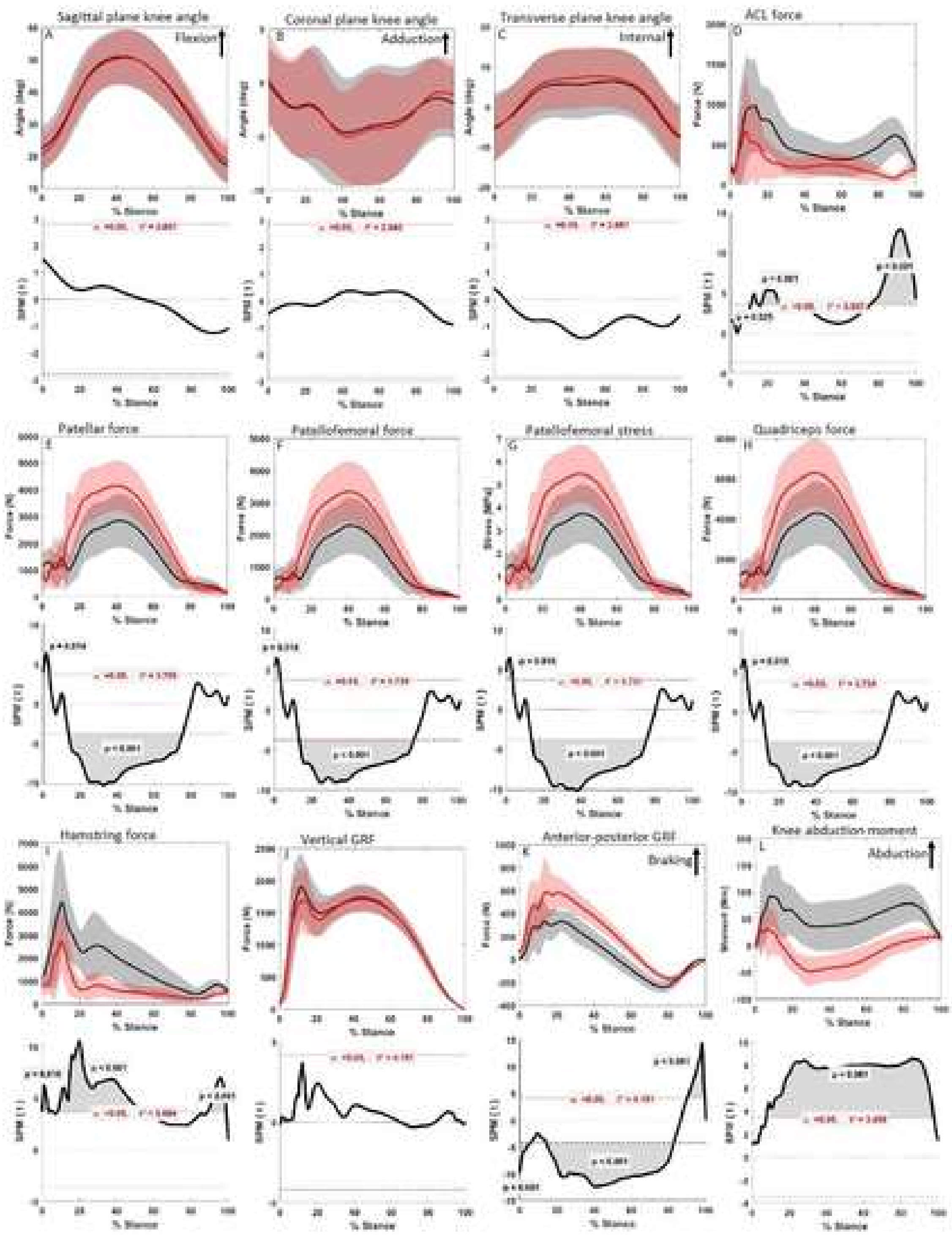


Figure

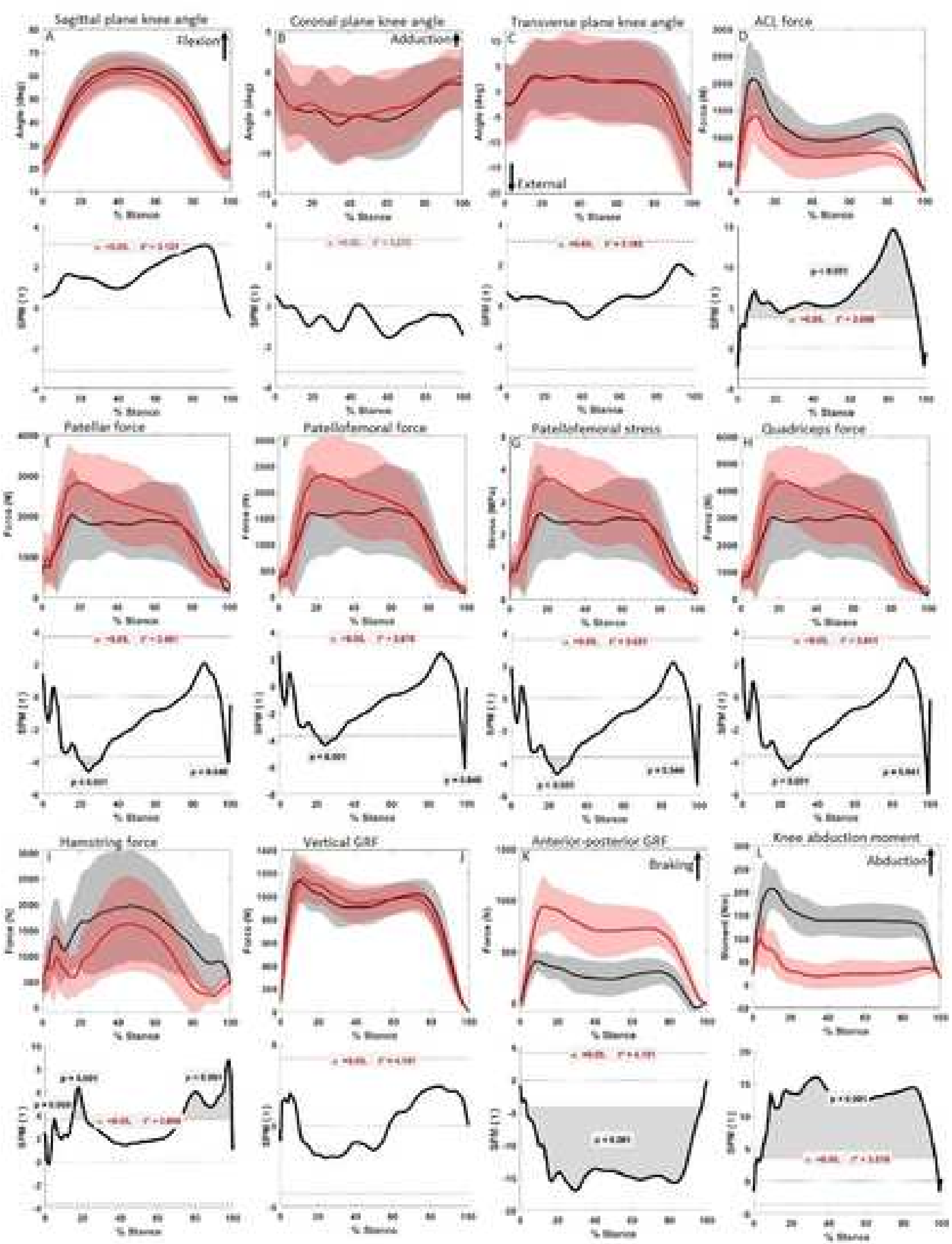
Click here to access/download;Figure;FIGURE 2 NEW.jpg



Figure


Sagittal plane knee angle
Coronal plane knee angle
Transverse plane knee angle
ACL force
Flexion
Adduction
Internal
Patellar force
Patellofemoral force
Patellofemoral stress
Quadriceps force
Hamstring force
Vertical GRF
Anterior-posterior GRF
Knee abduction moment
Braking
Abduction
% Stance

Figure


Sagittal plane knee angle
A
Flexion
Coronal plane knee angle
B
Adduction
Transverse plane knee angle
C
External
ACL force
D
Patellar force
E
Patellofemoral force
F
Patellofemoral stress
G
Quadriceps force
H
Hamstring force
I
Vertical GRF
J
Anterior-posterior GRF
K
Braking
Knee abduction moment
L
Abduction

Reviewer #1:

COMMENTS TO AUTHORS

Dear Authors,

I believe the topic of your article is interesting.

However, there are some points I would clarify, before publication.

I approve the publication of this paper after minor revision.

I have the following detailed comments:

TITLE

Ok

ABSTRACT

Well written

INTRODUCTION

The introduction provides adequate background.

METHODS

OK

RESULTS

OK

DISCUSSION/ CONCLUSION

It is recommend to reinforce the motivations and criteria that led you to the conclusions through this methodology and discuss why the applied method is appropriate.

RESPONSE: The mechanisms responsible for each conclusion is now added to each paragraph or relevance in the discussion.

With regards to the modelling methods with great respect, we feel that the efficacy of each model is already described in the methods section as part of the alterations included in the previous revision.

REFERENCES

Ok.

FIGURES

OK