Effects of prophylactic knee bracing on knee joint kinetics and kinematics during single and double limb post-catch deceleration strategies in university netballers.

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Abstract

PURPOSE: The aim of the current investigation was to investigate the effects of a prophylactic knee brace on knee joint kinetics and kinematics during single and double limb deceleration tasks. *METHODS:* Twenty female university first team level netballers performed single and double limb deceleration tasks under two conditions (prophylactic knee brace/ no-brace). Biomechanical data was captured using an eight-camera 3D motion capture system and a force platform. Participants also subjectively rated the comfort/ stability properties of the brace and their knee joint proprioception was examined with and without the knee brace using a weight bearing joint position sense test. *RESULTS:* The results showed that during both single and double limb deceleration tasks neither peak anterior cruciate ligament (brace: single=1.30 / double=1.30 bodyweight (BW) & no-brace: single=1.19 / double=1.29 BW) $P=0.51$, patellofemoral (brace: single=4.21/ double = 4.93 BW & no-brace: single=3.99 / double=4.63 BW) $P=0.20$ or patellar tendon (brace: single = 6.17/ double=6.49 BW & no-brace: single=6.07 / double=6.14 BW) $P=0.49$ kinetics were significantly affected as a function of wearing the knee brace. The findings also showed that the knee brace helped to increase participants perceived knee stability ($P<0.001$) but there were no statistical improvements in weight bearing knee proprioception (brace=3.59 & no-brace=2.94˚) $P=0.44$. *CONCLUSIONS:* The current investigation indicates that the utilization of prophylactic knee bracing akin to the device used in the current study does not appear to reduce the biomechanical parameters linked to the aetiology of knee injuries, during netball specific deceleration movements.

Introduction

Netball is representative of a team based global sporting discipline, with participation in over 70 countries (1). Like most court sports netball is a physical challenging activity characterized by a series of high intensity dynamic movements; although unlike most court based disciplines there are additional physical considerations imposed by the specific rules of the sport (2). Particularly as players must stop completely upon receiving the ball which places considerable emphasis on rapid deceleration manoeuvres (2).

Indeed, netball has been shown to be associated with a high rate of non-contact injuries. During tournament play 238 injuries were observed per 1000 playing hours (3) and an injury rate of 66.7–71.4 per 1000 participants has been noted from a retrospective analysis of three competitive seasons (4). These analyses have shown that the majority of injuries occur in the lower extremities; with the knee being the most commonly injured musculoskeletal structure in netball players, accounting for 24 % of total injuries (3, 5). Importantly, a systematic review of knee pathologies identified rapid deceleration manoeuvres as one of the three movements that may lead to knee injury (6).

Single and double limb landing manoeuvres generate large impact forces that are primarily attenuated in the lower extremities joints, with particular stress at the knee joint (7). The knee joint structures considered at greatest risk from chronic and acute pathologies during rapid deceleration tasks are the anterior cruciate ligament (ACL), patellofemoral joint and patellar tendon (8, 9, 10). Over 250,000 ACL injuries occur annually, causing long term absence from training (11), and allocated healthcare costs of over $3.4 billion (12). Biomechanically, the predominant risk factors for non-contact ACL pathologies are a reduced knee flexion angle at initial contact, large knee valgus angle, large knee internal rotation angle and excessive forces experienced by the ligament itself (11, 13, 14). In addition, patellofemoral pain accounts for a quarter of all injuries treated in sports medicine clinics, and is strongly linked to the aetiology of osteoarthritis at this joint (15). Kinetic and kinematic risk factors identified as predictors of future patellofemoral pain include, a decreased peak knee flexion angle, enhanced knee abduction, decreased vertical ground reaction force, elevated patellofemoral joint reaction force, and augmented patellofemoral joint stress (16). Similarly, chronic patellar tendinopathy (or jumper's knee) may account for up to 25 % of all soft tissue injuries, and forces 53 % of symptomatic athletes to permanently cease physical activities (17). Biomechanical risk factors linked to the aetiology of patellar tendinopathy include decreased knee flexion, knee flexion range of motion (ROM), increased patellar tendon force and higher patellar tendon rate of loading (9, 18).

Knee braces are commonly utilized in high intensity activities sports such as netball in order to prevent knee injuries and improve symptoms in those with existing pathologies (19). Knee braces represent external devices which are designed in order to positively influence the position of the patella relative to the trochlear groove and improve knee alignment (20). They range from fixed devices which typically include uniaxial or polyaxial vertical hinges to more compliant knee sleeves designed to provide knee compression and improve proprioception (21). Knee braces are a low cost conservative modality that can be utilized during sports manoeuvres (22). Prophylactic knee braces are designed in order to prevent sportspersons knee injuries whilst also being minimally restrictive, although there is currently little published evidence to support their effectiveness in shielding the knee from injury (22).

The effects of knee bracing have been studied extensively in a range of sports movements. However, there is currently only one investigation which has examined the effects of knee bracing in netball players. Sinclair et al., (18) examined the influence of a prophylactic knee brace on patellofemoral joint kinetics and three-dimensional knee joint kinematics during run, cut and vertical jump movements. Their findings confirmed that there were no differences in patellofemoral joint kinetics as a function of wearing the knee brace, but knee joint range of motion in the transverse plane was statistically attenuated. However, there is yet to be any published information concerning the effects of knee bracing in netball players during single and double limb deceleration tasks.

Therefore the aim of the current investigation was to investigate the effects of a prophylactic knee brace on knee joint kinetics and kinematics during single and double limb deceleration tasks. The findings may provide both coaches and netballers with information regarding the utilization of knee bracing for the attenuation of the biomechanical parameters linked to the aetiology of knee injuries during high intensity netball specific movements.

Methods

Participants

Twenty female netball players (age = 19.92 ± 0.79 years, height = 1.66 ± 0.05 m, mass = 62.43 ± 8.66 kg) were recruited to for this study. This sample size is commensurate with previous analyses concerning the effects of prophylactic bracing on knee joint kinetics and kinematics in netball specific movements (19). Volunteers were considered eligible for participation if they were; over 18, university first team level players and possessed a minimum of 3 years of competitive netball experience. Participants were excluded from the study if there was evidence of existing knee pathology or there had been previous knee surgery. Written informed consent was provided and the procedure was approved by the University.

Knee Brace

A single nylon/silicone knee brace was utilized in this investigation, (Kuangmi 1 PC compression knee sleeve), which was worn on the dominant (right) limb in all participants. The brace examined as part of this study is lightweight knee joint compression sleeve designed to provide support and enhance joint proprioception.

Procedure

Participants were required to complete five repetitions of a simulated centre pass movement (described below), with and without presence of the brace. The order that participants performed in the movement/ brace conditions was counterbalanced. For the single limb movement condition, participants were instructed to jog towards the force platform, when they were within 0.75 m of the plate (marked using masking tape) a regulation size netball (Gilbert Spectra, Size 5) was passed to them in the opposing direction that they were moving, by a single university 1st team level netball player. Having caught the ball participants were required to decelerate by planting their dominant (right) limb on the force platform prior to the contralateral side. For the double limb condition the process was identical but participants were required to land with both feet simultaneously, with only the dominant limb on the force plate. Participants were allowed as much practice time/trials to accommodate to the experimental conditions as they deemed necessary. To ensure that participants utilized a similar approach velocity in the brace and no-brace conditions; the linear velocity of the pelvic segment was quantified. The approach velocity during the first trial in both the single and double limb movement conditions was calculated and a maximum deviation of 5% from this velocity was allowed throughout data collection for each participant (23). Both movements were defined as the duration from foot contact (defined as > 20N of vertical force applied to the force platform), to maximum knee flexion (19).

In addition to the biomechanical movement information, the effects of the experimental brace on knee joint proprioception were also examined using a weight bearing joint position sense test. This was conducted, in accordance with the procedure of Drouin, et al., (24), whereby participants were assessed on their ability to reproduce a target knee flexion angle of 30˚ whilst in single leg stance. To accomplish this, participants were asked to slowly squat to a knee flexion angle of 30 ˚, which was verified using a handheld goniometer by the same researcher throughout data collection. Participants then held this position for 15 seconds during which time the knee criterion angle was captured using the motion analysis system. Following this participants were asked to return to a standing position and wait for 15 seconds, and they were required to repeat the above process without guidance via the goniometer. Again this position was held for a period of 15 seconds and the replication trial was also collected using the motion analysis system. This above process conducted on three occasions in both the brace and no-brace conditions in a counterbalanced order, and between each trial participants walked for 20 ft to eliminate any proprioceptive memory of the previous trial. The absolute difference in degrees calculated between the criterion and replication trials was averaged over the three trials to provide an angular error value in both brace and no-brace conditions, which was extracted for statistical analysis.

Kinematics and ground reaction force (GRF) information were synchronously collected. Kinematic data were captured at 250 Hz via an eight camera motion analysis system

(Qualisys Medical AB, Goteburg, Sweden) and kinetic data using a force platform (Kistler, Kistler Instruments Ltd., Alton, Hampshire) which operated at 1000 Hz. Dynamic calibration of the motion capture system was performed before each data collection session. To quantify lower extremity segments in six degrees of freedom, the calibrated anatomical systems technique was utilized. To define the anatomical frames of the pelvis, thigh, shank and foot retroreflective markers (19 mm) were positioned onto the, iliac crest, anterior superior iliac spine (ASIS), and posterior super iliac spine (PSIS). In addition, further markers were placed unilaterally onto the, medial and lateral malleoli, greater trochanter, medial and lateral femoral epicondyles calcaneus, first metatarsal and fifth metatarsal heads of the affected limb. Carbon-fiber tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. In addition to these the foot segments were tracked via the calcaneus, first metatarsal and fifth metatarsal, and the pelvic segment was tracked using the PSIS and ASIS markers. The hip joint centre was determined using a regression equation, which uses the positions of the ASIS markers and the centers of the ankle and knee joints were delineated as the mid-point between the malleoli and femoral epicondyle markers.

Static calibration trials were obtained with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters/markers. A static trial was conducted with the participant in the anatomical position in order for the anatomical positions to be referenced in relation to the tracking markers, following which those not required for dynamic data were removed. 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.

Following completion of the biomechanical data collection, in accordance with Sinclair et al., (19); participants were asked to subjectively rate the knee sleeve in relation to performing the movements without the brace in terms of stability and comfort. This was accomplished using 3 point scales that ranged from 1 = more comfortable, 2 = no-change and 3 = less comfortable and 1 = more stable, 2 = no-change and 3 = less stable. In addition, each participant was asked whether they would or would not choose to wear the knee brace during their training/ competitive netball activities.

Data 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 normalized to 100 % of the landing 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 (19). Three dimensional kinematics of the knee and ankle were calculated using an XYZ cardan sequence of rotations (where X = sagittal plane; Y = coronal plane and Z = transverse plane). Three dimensional knee joint angular kinematic measures that were extracted for statistical analysis were 1) angle at footstrike, 2) peak angle and 3) angular ROM from footstrike to peak angle.

Patellofemoral loading during the stance phase of running was quantified using a model adapted from van Eijden et al., (25), in accordance with the protocol of Willson et al., (26). A drawback of the van Eijden model is that co-contraction of the knee flexor musculature is not accounted for (26). In order to account for this, we also calculated hamstring and gastrocnemius forces in accordance with the procedures described by DeVita & Hortobagyi, (27). To summarize, the hamstring force was calculated using the hip extensor moment, hamstrings and gluteus maximus cross-sectional areas (28) and by fitting a 2nd order polynomial curve to the data of Nemeth & Ohlsen, (29) who provided muscle moment arms at the hip as a function of hip flexion angle. The gastrocnemius force was calculated firstly by quantifying the ankle plantarflexor force, which was resolved by dividing the plantarflexion moment by the Achilles tendon moment arm. The Achilles tendon moment arm was calculated by fitting a 2nd order polynomial curve to the ankle plantarflexion angle in accordance with Self & Paine, (30). Plantarflexion force accredited to the gastrocnemius muscles was calculated via the cross-sectional area of this muscle relative to the triceps surae (28).

The hamstring and gastrocnemius forces were multiplied by their estimated muscle moment arms to the knee joint in relation to the knee flexion angle (31), and then added together to estimate the knee flexor moment. The derived knee flexor moment was added to the net knee extensor moment quantified using inverse dynamics were then summed and subsequently divided by the quadriceps muscle moment arm (25), to obtain quadriceps force adjusted for co-contraction of the knee flexor musculature. Patellofemoral force was then quantified by multiplying the adjusted quadriceps force by a constant which was obtained by using the data of van Eijden et al., (25).

Finally, patellofemoral joint stress was quantified by dividing the patellofemoral force by the patellofemoral contact area. Patellofemoral contact areas were obtained by fitting a 2^{nd} order polynomial curve to the sex specific data of Besier et al., (32), who estimated patellofemoral contact areas as a function of the knee flexion angle using MRI. All patellofemoral forces were normalized by dividing the net values by bodyweight (BW). From the above processing, peak patellofemoral force, and peak patellofemoral stress (KPa/BW) were extracted. Patellofemoral instantaneous load rate (BW/s) was also extracted by obtaining the peak increase in force between adjacent data points.

In addition, Patellar tendon loading was quantified using a model similarly adapted from Janssen et al., (9). 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 tendon moment arm was quantified as a function of the sagittal plane knee angle, by fitting a 2^{nd} order polynomial curve to the data provided by Herzog & Read, (33). All patellar tendon forces were normalized by dividing the net values by bodyweight (BW). From the above processing, peak patellar tendon force was extracted. Patellar tendon instantaneous load rate (BW/s) was also extracted by obtaining the peak increase in force between adjacent data points.

Finally, ACL loading was quantified using the model described previously by Sinclair & Stainton, (23). All ACL forces were normalized by dividing the net values by bodyweight (BW). From the above processing, peak ACL force was extracted. ACL instantaneous load rate (BW/s) was also extracted by obtaining the peak increase in force between adjacent data points.

Statistical analyses

Descriptive statistics of means and standard deviations were obtained for each outcome measure. Shapiro-Wilk tests were used to screen the data for normality. Differences in knee proprioception with and without the presence of the brace were examined using the using a paired t-test. Differences in biomechanical and knee pain parameters were examined using 2 (BRACE) x 2 (MOVEMENT) repeated measures ANOVA's. Statistical significance was accepted at the $P \leq 0.05$ level. Effect sizes for all significant findings were calculated using partial Eta^2 ($p\eta^2$). All statistical actions were conducted using SPSS v24.0 (SPSS Inc, Chicago, USA).

Results

Tables 1-3 present the mean ± SD knee kinetics and kinematics as a function of different brace and movement conditions. Figure 1 shows the mean ± SD knee proprioception as a function of wearing the knee brace.

Patellofemoral loading

A significant main effect of MOVEMENT ($P<0.05$, $p\eta^2 = 0.43$) was noted for peak patellofemoral load, with the highest forces being experienced in the double limb landing (Table 1). A significant main effect of movement ($P<0.05$, $p\eta^2 = 0.41$) was also revealed noted for the patellofemoral load rate, with the highest rates of loading being experienced in the double limb landing (Table 1).

Patellar tendon loading

No significant ($P>0.05$) differences were observed for patellar tendon loading (Table 1).

@@@TABLE 1 NEAR HERE@@@

ACL loading and muscle kinetics

No significant ($P>0.05$) differences were observed for ACL loading (Table 2).

@@@TABLE 2 NEAR HERE@@@

Three-dimensional kinematics

In the sagittal plane a significant main effect of MOVEMENT ($P<0.05$, $p\eta^2 = 0.69$) was noted for the knee flexion angle at footstrike, which was greater in the double limb landing condition (Table 3). In addition, for peak knee flexion there were significant main effects for both MOVEMENT ($P<0.05$, $p\eta^2 = 0.39$) and BRACE ($P<0.05$, $p\eta^2 = 0.62$). Peak flexion was found to be greater in the double limb landing and also in the brace condition (Table 3). Finally, for sagittal ROM there was a main effects of BRACE ($P<0.05$, $p\eta^2 = 0.37$), which was found to be greater in the brace condition (Table 3).

In the coronal plane a significant main effect of MOVEMENT ($P<0.05$, $p\eta^2 = 0.36$) was noted for the knee abduction angle at footstrike, which was greater in the double limb landing condition (Table 3). In addition there was also a main effect of MOVEMENT ($P<0.05$, $p\eta^2 = 0.37$), for the peak knee abduction angle, which was shown to be greater in the double leg landing condition (Table 3). Finally, for coronal plane ROM there was a main effects of movement ($P<0.05$, $p\eta^2 = 0.48$), which was found to be greater in the double leg landing condition (Table 3).

In the transverse plane a significant main effect of BRACE ($P<0.05$, $p\eta^2 = 0.37$), was noted for the knee external rotation angle at footstrike, which was significantly lower in the brace condition (Table 3).

@@@TABLE 3 NEAR HERE@@@

Knee proprioception

No significant ($P=0.44$) differences in knee proprioception were observed.

@@@FIGURE 1 NEAR HERE@@@

Subjective ratings

Subjective ratings of comfort showed no significant changes were found when wearing the knee braces (X^2=0.70, P=0.40), with 5 participants rating the brace as more comfortable, 7 no change and 8 less comfortable. However, participants subjectively rated that wearing the knee brace significantly increased stability during both landings (X^2= 14.80, P<0.001), with 14 participants rating the brace as more stable, 6 no change and 0 less stable. Finally, no significant change was observed for participants subjective indication of whether they would choose to wear the brace (X^2= 1.80, P=0.18), with 7 participants indicating that they would wear the brace for their netball training/ competition activities and 13 indicating that they would not.

Discussion

To the authors knowledge this represents the first investigation to explore the influence of prophylactic knee bracing during netball specific deceleration tasks and thus may provide important information to netballers and clinicians regarding the efficacy of knee bracing in this sporting discipline. The findings from this study show that whilst participants perceived that the brace significantly improved joint stability, the presence of the brace did not mediate any significant alterations in the kinetic/ kinematic parameters linked to the aetiology of injury.

The current investigation showed firstly that neither ACL, patellofemoral or patellar tendon loading were statistically influenced as a function of the knee brace condition. This observation is in agreement with those of Sinclair et al., (19) who showed that knee bracing did not significantly affect patellofemoral loading during netball specific movements, although it should be noted that neither ACL or patellar tendon kinetics were examined in this study. As the current study utilized a lightweight nylon/ silicone construction, it is proposed that this observation relates to the mechanical structure of the knee brace which was not able to provide sufficient physical restraint to mediate alterations in knee joint loading. Nonetheless excessive loading at the ACL, patellofemoral joint and patellar tendon are considered to be one of the key mechanisms linked to the aetiology of knee pathologies in athletic populations (9, 12, 16). Therefore the key implication from this observation is that the prophylactic brace examined in this study does not appear to reduce the knee kinetic parameters that have been linked to the aetiology of knee pathologies in netball specific single and double limb deceleration tasks.

It has been proposed that knee bracing facilitates safer movement mechanics during dynamic activities, by promoting an enhanced perception of joint stability (34). The subjective ratings of stability noted in the current investigation support this notion in that participants perceived that the knee brace significantly improved knee joint stability. However, the current investigation also showed that knee proprioception was not statistically improved as a function of wearing the prophylactic knee brace. This indicates that the perceived change in stability was not apparent in either the deceleration movements or the proprioceptive task. Proprioceptive acuteness, an element of the sensorimotor system, is reflective of an athlete's ability to perceive joint position, motion and external forces in order to differentiate lower limb movement (35). As such, improving knee joint proprioception acuity is considered an essential component for injury prevention as it is makes the knee joint more receptive to potentially injurious forces (36).

The observations from this investigation concur with those of Bottoni, et al., (37) yet disagree with the observations of Birmingham et al., (38), Herrington et al., (34) and Van Tiggelen et al., (39). The lack of agreement between studies in general is due to the lack of standardization of testing protocols to quantify knee joint proprioception (33). However, the current investigation selected a weight bearing joint position sense protocol based on the notion proposed by Hanafy, (40), that this technique provides more clinical and ecological relevance when evaluating proprioception in relation to weight bearing specific pathologies. Nonetheless the current investigation has demonstrated that prophylactic knee bracing does not improve knee joint proprioception in a weight bearing angle reproduction test in netball players. The proposed mechanism by which knee bracing is considered to enhance joint proprioception is through compression of the skin/ musculature, which serves to stimulate sense receptors and increase the afferent input from the joint surrounding structures (34). Thus is can be speculated that the brace may not have provided sufficient compression to the knee to mediate statistical improvements in joint proprioception. Further research into the association between compression provided by the knee brace and joint proprioception is thus a clear avenue for further investigation.

A potential limitation to this work is that joint kinetics were obtained using a musculoskeletal modelling approach as opposed to an in vivo exploration of knee loading. This process was necessary due to the impracticalities and invasive nature of obtaining direct kinetic measurements. However, although this approach represents expansion compared to previous mechanisms in that co-contraction of the knee flexor musculature was accounted for, further work is required to improve the efficacy of subject specific knee joint musculoskeletal models which will make possible further developments in clinical biomechanics. In addition, a further potential limitation of the current investigation is that it examines healthy netballers who did not habitually wear knee bracing. This means that the findings are not generalizable to netballers with existing knee joint pathology. Future, prospective analyses will help to determine the clinical efficacy of knee braces as treatment modalities for netballers with existing knee injuries.

Conclusion

This study showed firstly that neither ACL, patellofemoral nor patellar tendon kinetic parameters were significantly affected as a function of the knee brace. The findings did show however that the knee brace helped to increase perceived knee stability, but there were no statistical improvements in weight bearing knee proprioception. This indicates that the perceived change in stability was not apparent in either the deceleration movements or proprioceptive tasks. The current investigation indicates that the utilization of prophylactic knee bracing akin to the device used in the current study, does not appear to reduce the biomechanical parameters linked to the aetiology of knee injuries, during netball specific deceleration movements. However, further prospective analyses are required to fully substantiate this proposition.

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Conflict statement

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Tables

Table 1: Mean ± SD patellofemoral and patellar tendon kinetics as a function of the knee brace and different movement conditions.

	No-brace				Brace				P-value		
	Single		Double		Single		Double				
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	BRACE	MOVEMENT	*BRACE * MOVEMENT*
Peak patellofemoral force (BW)	3.99	0.98	4.63	1.33	4.21	1.57	4.93	1.37	0.20	0.02	0.84
Peak patellofemoral stress (KPa/BW)	15.12	2.82	15.21	3.85	15.15	4.35	16.11	3.43	0.42	0.39	0.50
Patellofemoral load rate (BW/s)	119.82	24.76	144.11	50.96	108.82	36.34	137.08	42.55	0.21	0.02	0.80
Peak patellar tendon force (BW)	6.07	1.23	6.14	1.56	6.17	1.75	6.49	1.42	0.49	0.32	0.63
Patellar tendon load rate (BW/s)	246.31	50.38	281.14	92.09	219.89	77.69	263.19	81.68	0.07	0.14	0.75

Table 2: Mean ± SD ACL kinetics as a function of the knee brace and different movement conditions.

	No-brace				Brace				P-value		
	Single		Double		Single		Double				
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	BRACE	MOVEMENT	*BRACE * MOVEMENT*
Peak ACL force (BW)	1.19	0.38	1.29	0.31	1.30	0.39	1.30	0.37	0.51	0.52	0.56
ACL load rate (BW/s)	113.59	51.69	115.04	53.56	131.49	57.53	106.12	35.03	0.11	0.69	0.12

Table 3: Mean ± SD knee joint kinematics as a function of the knee brace and different movement conditions.

	No-brace				Brace				P-value		
	Single		Double		Single		Double				
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	BRACE	MOVEMENT	*BRACE * MOVEMENT*
Sagittal plane (positive = flexion)											
Angle at footstrike (°)	16.35	3.73	20.54	5.14	17.70	4.77	22.18	6.85	<0.001	0.11	0.80
Peak flexion (°)	60.37	7.43	69.91	8.82	65.30	9.21	72.72	11.29	0.02	0.001	0.37
ROM (°)	44.02	6.72	49.37	9.02	47.60	9.08	50.54	9.39	0.17	0.03	0.35
Coronal plane (positive = abduction)											
Angle at footstrike (°)	0.93	4.08	0.64	4.31	1.92	3.82	0.89	3.48	0.04	0.34	0.11
Peak abduction (°)	5.20	7.26	8.64	8.22	7.02	8.10	9.42	7.63	0.03	0.29	0.38
ROM (°)	4.27	4.00	8.00	5.63	5.10	5.79	8.53	5.52	0.009	0.56	0.80
Transverse plane (positive = external rotation)											
Angle at footstrike (°)	9.57	11.71	8.83	8.47	6.05	10.41	6.24	9.57	0.81	0.03	0.56
Peak external rotation (°)	-4.67	8.53	-5.66	6.79	-6.90	8.87	-7.13	6.81	0.49	0.08	0.28
ROM (°)	14.25	4.50	14.49	4.35	12.95	5.22	13.36	6.24	0.73	0.33	0.91

Figure labels

Figure 1: Mean ± SD angular error values for both brace and no-brace conditions during the weight bearing joint position sense test.