Effects of prophylactic knee bracing on patellar tendon loading parameters during functional sports tasks in recreational athletes

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Abstract

PURPOSE: This study investigated the effects of prophylactic knee bracing on patellar tendon loading parameters.

METHODS: Twenty recreational athletes (10 male & 10 female), from a different athletic disciplines performed run, cut and single leg hop movements under two conditions (prophylactic knee brace/ no-brace). Lower extremity kinetics and kinematics were examined using a piezoelectric force plate, and three-dimensional motion capture system. Patellar tendon loading was explored using a mathematical modelling approach, which accounted for co-contraction of the knee flexors. Tendon loading parameters were examined using 2 (*brace*)*3 (*movement*)*2 (*sex*) mixed ANOVA's.

RESULTS: Tendon instantaneous load rate was significantly reduced in female athletes, in the run (brace = 289.14BW/s no-brace = 370.06BW/s) and cut (brace = 353.17BW/s/ no-brace = 422.01BW/s) conditions whilst wearing the brace.

CONCLUSIONS: Female athletes may be able to attenuate their risk from patellar tendinopathy during athletic movements, through utilization of knee bracing, although further prospective research into the prophylactic effects of knee bracing is required before this can be clinically substantiated.

Introduction

Chronic patellar tendinopathy is an extremely common musculoskeletal condition in both recreational and elite athletes, and has previously been reported to account for as many as 25% of all soft tissue injuries (1). Patellar tendinopathy is characterized by pain localized at the lower pole of the patella, and pain symptoms that are augmented by activities which place high demands on the knee extensors, notably in physical disciplines which repeatedly store and release elastic energy in the tendon itself (2). Patellar tendinopathy is more common in skeletally mature individuals, and there remains disagreement as to whether this condition is most common in male or female athletes (3). Chronic patellar tendinopathy is established after 1-3 months, as degenerative alterations occur in the tendon itself (4). Degenerative alterations at the tendon are mediated primarily by the absence of inflammatory cells within the tendon itself, which reduces healing of the tendon and ultimately leads to decreased tensile strength and disorganization of the collagen fibers (5). Patellar tendinopathy can be debilitating; Cook et al., (6) showed that 1/3 of athletes with patellar tendinopathy are unable to return to physical activity within 6 months, and it has also been evidenced that 53% of athletes who present with this condition were forced to permanently cease physical activities.

Knee braces are utilized extensively in both recreationally active and competitive athletes, in order to attenuate their risk from knee pathology (7). Knee braces are external devices which are designed to improve the alignment of the knee joint (8). Prophylactic knee braces aim to protect athletes from sustaining injury, whilst being minimally restrictive, allowing athletes to utilize full knee range of motion during their physical activities (9). Recently, the effects of prophylactic knee braces on the biomechanics of the knee joint during dynamic sports tasks have received significant attention in clinical literature. Sinclair et al., (7), examined the effects of knee bracing on knee joint kinetics and kinematics in netball specific movements. They showed that the brace did not alter knee kinetics but did reduce range of motion in the transverse plane. Ewing et al., (10), examined muscle kinetics with and without the presence of a prophylactic knee brace during double limb drop landings. Hamstring and vasti muscles produced significantly greater flexion and extension torques, and greater peak muscle forces in the brace condition. Lee et al., (11), analyzed the effects of a prophylactic bilateral hinge brace, fitted with torque transducers during four functional sports tasks; drop vertical jump, pivot, stop vertical jump and cut. Their results showed that the knee brace hinges absorbed up to 18% of the force and 2.7% of the torque at the knee, during the different athletic motions. Which they concluded, was minimal evidence that the brace was able to reduce the mechanical load at the knee. Although knee braces have been studied in terms of both their therapeutic and prophylactic effects, there is currently no literature which has considered their role in the prevention of patellar tendinopathy.

Therefore, the aim of the current investigation was to investigate the effects of a prophylactic knee brace on patellar tendon loading parameters linked to the aetiology of patellar tendinopathy, in male and female recreational athletes. Research of this nature may provide important clinical information, regarding the potential role of prophylactic knee bracing for the prevention of patellar tendinopathy.

Methods

Participants

Twenty participants (10 male; age = 26.70 ± 4.24, mass = 73.90 ± 5.3, stature = 176.50 ± 4.25 & BMI = 23.73 ± 1.80 & and 10 female age = 27.60 ± 4.72, mass = 60.40 ± 7.86, stature = 166.50 ± 5.06 & BMI = 21.86 ± 2.21), volunteered to take part in the current investigation. Participants were all recreational level athletes who came from squash, netball, basketball and association football athletic backgrounds, with a minimum of 2 years of experience in their chosen discipline. In addition, all were free from lower extremity pathology at the time of data collection, and had not previously experienced an injury to the patellar tendon. Written informed consent was provide,d in accordance with the declaration of Helsinki and the rights of all participants were protected. The procedure was approved by the Universities Science, Technology, Engineering, Medicine and Health ethics committee, with the reference STEMH 295.

Knee Brace

A single knee brace was utilized in this investigation, (Trizone, DJO USA), which was worn on the dominant limb in all participants. The brace examined in the current investigation represents a compression sleeve reinforced with silicone designed to support the knee joint and improve proprioception.

Procedure

Participants were required to complete five repetitions of three sports specific movements'; jog, cut and single leg hop, with and without presence of the brace. The order that participants performed in the movement/ brace conditions was counterbalanced. To quantify lower extremity segments, the calibrated anatomical systems technique was utilized (12). Retroreflective markers (19 mm), were positioned unilaterally allowing the; foot, shank and thigh to be defined. The foot was defined via the 1st and 5th metatarsal heads, medial and lateral malleoli and tracked using the calcaneus, 1st metatarsal and 5th metatarsal heads. The shank was defined via the medial and lateral malleoli and medial and lateral femoral epicondyles and tracked using a cluster positioned onto the shank. The thigh was defined via the medial and lateral femoral epicondyles and the hip joint centre and tracked using a cluster positioned onto the thigh. To define the pelvis additional markers were positioned onto the anterior (ASIS) and posterior (PSIS) superior iliac spines and this segment was tracked using the same markers. The hip joint centre was determined using a regression equation, which uses the positions of the ASIS markers (13). The centers of the ankle and knee joints were delineated as the mid-point between the malleoli and femoral epicondyle markers (14, 15). Each tracking cluster comprised four retroreflective markers, mounted onto a rigid piece of lightweight carbon-fibre. Static calibration trials were obtained 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.

Data were collected during run, cut and jump movements using the protocol below:

Run

Participants ran at 4.0 $m.s^{-1}$ ±5%, and struck the force platform with their right (dominant) limb. The average velocity of running was monitored using infra-red timing gates (SmartSpeed Ltd UK). The stance phase of running, was defined as the duration over > 20 N of vertical force was applied to the force platform (16).

Cut

Participants completed 45° sideways cut movements, using an approach velocity of 4.0 $m.s^{-1}$ ±5% striking the force platform with their right (dominant) limb. In accordance with McLean et al., (17), 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 the cut-movement was similarly defined as the duration over > 20 N of vertical force was applied to the force platform (16).

Hop

Participants began standing by on their dominant limb; they were then requested to hop forward maximally, landing on the force platform with same leg without losing balance. The arms were held across the chest to remove arm-swing contribution. The hop movement was defined as the duration from foot contact (defined as > 20 N of vertical force applied to the force platform), to maximum knee flexion. The hop distance was recorded and maintained throughout data collection.

Processing

Dynamic trials were processed using Qualisys Track Manager, and then exported as C3D files. Ground reaction force and marker data were filtered at 50 Hz and 15 Hz respectively using a low-pass Butterworth 4th order filter, and processed using Visual 3-D (C-Motion, Germantown, MD, USA). Internal moments were computed using Newton-Euler inverse-dynamics, allowing net knee joint moments to be calculated. Angular kinematics of the knee joint were calculated using an XYZ (sagittal, coronal and transverse) sequence of rotations, allowing sagittal angles at footstrike and peak flexion angles to be extracted.

A commonly utilized mathematical model for the quantification of patellar tendon loading is that developed by Janssen et al., (18). Whereby the Patellar tendon load is determined by dividing the knee extensor moment by the estimated patellar tendon moment arm. This algorithm has been successfully utilized previously, to resolve differences in patellar tendon kinetics during different movements (18), different footwear conditions (19), and also between sexes (20).

However, a limitation of the aforementioned model is that the knee extensor moment does not account for co-contraction of the knee flexor musculature. In order to account for this, we also calculated hamstring and gastrocnemius force in accordance with the procedures described by DeVita and Hortobagyi (21). To summarize, the hamstring force was calculated using the hip extensor moment, hamstrings and gluteus maximus cross-sectional areas (22), and by fitting a 2nd order polynomial curve to the data of Nemeth & Ohlsen, (23) 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 and Paine (24). The quantity of plantarflexion force accredited to the gastrocnemius muscles, was calculated via the cross-sectional area of this muscle relative to the triceps surae (22).

The hamstring and gastrocnemius forces were multiplied by their estimated muscle moment arms to the knee joint in relation to the knee flexion angle (25), 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, 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 2nd order polynomial curve to the data provided by Herzog & Read, (26), showing patellar tendon moment arms at different knee flexion angles.

All patellar tendon load parameters were normalized by dividing the net values by bodyweight (BW). Patellar tendon instantaneous load rate (BW/s), was quantified as the peak increase in patellar tendon force between adjacent data points. In addition, we also calculated the total patellar tendon force impulse (BW·s) during each movement using a trapezoidal function.

Statistical analyses

Descriptive statistics of means, standard deviations and 95% confidence intervals (95% CI) were obtained for each outcome measure. Shapiro-Wilk tests were used to screen the data for normality. Differences in patellar tendon loading parameters between conditions, were examined using 2 (*brace*) * 3 (*movement*) * 2 (*sex*) mixed ANOVA's. Statistical significance was accepted at the $P<0.05$ level. Effect sizes for all significant findings were calculated using partial Eta^2 (pη^2). Post-hoc pairwise comparisons were conducted on all significant main effects. Significant interactions were further evaluated by performing simple main effect examinations on each level of the interaction, in the event of a significant simple main effect pairwise comparisons were performed. All statistical actions were conducted using SPSS v22.0 (SPSS Inc, Chicago, USA).

Results

Tables 1-4 and figure 1 present patellar tendon loading parameters as a function of *brace*, *movement* and *sex*.

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Peak patellar tendon force

A significant main effect ($P<.05$, $p\eta^2 = .20$) was found for *movement*. Post-hoc pairwise comparisons showed that peak patellar tendon force was significantly larger in the cut movement compared to the hop ($P=.046$) and run ($P=.008$) conditions.

In addition a significant main effect ($P<.05$, $p\eta^2 = .31$) was observed for *brace*. Post-hoc pairwise comparisons showed that peak patellar tendon force was significantly larger in the no-brace ($P=.013$) condition compared to wearing the brace.

Patellar tendon instantaneous load rate

A significant main effect ($P<.05$, $p\eta^2 = .29$) was found for *movement*. Post-hoc pairwise comparisons showed that patellar tendon instantaneous load rate was significantly larger in the cut ($P=.032$) and hop ($P=.003$) conditions compared to the run movement. In addition a significant main effect ($P<.05$, $p\eta^2 = .45$) was observed for *brace*, with patellar tendon instantaneous load rate being significantly in the no-brace condition compared to wearing the brace.

Finally a significant ($P<.05$, $p\eta^2 = .19$) *brace* * *movement* * *sex* interaction was shown. Follow up analyses using simple main effects showed for males that a there was a significant main effect ($P<.05$, $p\eta^2 = .21$) for *movement*, with the hop ($P=.01$) and cut ($P=.04$)

movements being associated with a greater instantaneous load rate than the run movement. For females there was a significant main effect (P<.05, $p\eta^2 = .86$) for *movement*, with the hop (P=.00001) and cut (P=.002) movements being associated with a greater instantaneous load rate than the run movement. In addition there was also a main effect (P<.05, $p\eta^2 = .57$) for *brace* with instantaneous load rate being significantly (P=.018) larger in the no-brace condition. Finally a significant (P<.05, $p\eta^2 = .42$) *brace * movement* interaction was found for females. Follow up analyses showed that there were main effects for the run (P<.05, $p\eta^2 = .89$) and cut (P<.05, $p\eta^2 = .72$) movements, with instantaneous load rate being significantly greater in the no-brace condition for both movements (cut – P=.004 & run – P=.00001). No differences were shown for the hop condition.

Patellar tendon impulse

A significant main effect (P<.05, $p\eta^2 = .20$) was found for *movement*. Post-hoc pairwise comparisons showed that peak tendon impulse was significantly larger in the cut (P=.0002) and hop (P=.048) movements compared to the run condition.

In addition a significant main effect (P<.05, $p\eta^2 = .19$) was observed for *brace*, with patellar tendon impulse was significantly larger in the no-brace (P=.042) condition compared to wearing the brace.

Finally, a significant (P<.05, $p\eta^2 = .19$) *brace * movement * sex* interaction was shown. Follow up analyses using simple main effects showed for males that a there was a significant main effect (P<.05, $p\eta^2 = .35$) for *movement*, with the hop (P=.001) and cut (P=.023) movements being associated with a greater impulse than the run movement. For females there was a significant main effect (P<.05, $p\eta^2 = .22$) for *movement*, with the cut (P=.01) being associated with a greater impulse than the run movement. Finally a significant ($P<.05$, $p\eta^2 = .56$) *brace * movement* interaction was found for females. Follow up analyses showed that there was a main effect for the run ($P<.05$, $p\eta^2 = .89$) movement, with impulse being significantly ($P=.0004$) greater in the no-brace condition.

Sagittal knee kinematics

For the knee flexion angle at footstrike, a significant main effect ($P<.05$, $p\eta^2 = .36$) was observed for *brace,* with knee flexion being reduced in the brace condition. For the peak flexion angle, a significant main effect ($P<.05$, $p\eta^2 = .28$) was observed for *brace*, with peak flexion being reduced in the brace condition. In, addition, a significant main effect ($P<.05$, $p\eta^2 = .60$) was observed for *movement.* Post-hoc pairwise comparisons indicated that peak flexion was significantly greater in the cut ($P=.000008$) and hop ($P=.0000009$) movement in comparison to the run and also in the hop compared to the cut ($P=.02$). Finally, a significant *brace * sex* ($P<.05$, $p\eta^2 = .22$) interaction was found. Follow up analyses showed that in female athletes only peak knee flexion was significantly reduced in the brace condition for the run ($P<.05$, $p\eta^2 = .37$) and hop ($P<.05$, $p\eta^2 = .66$) movements.

Discussion

The aim of the current investigation was to investigate the effects of a prophylactic knee brace on patellar tendon loading parameters linked to the aetiology of patellar tendinopathy, in male and female recreational athletes. To the authors’ knowledge, this represents the first investigation to examine the effects of prophylactic knee bracing in relation to the aetiology patellar tendinopathy.

A key finding from the current study is that indices of patellar tendon instantaneous load rate and impulse were found to be significantly reduced in female athletes during the run and cut movements when wearing the knee brace. This observation is interesting in that female athletes exhibited significant reductions in patellar tendon loading parameters as a function of the prophylactic brace, yet in male athletes there were no statistical alterations. The mechanisms responsible for this observation are unknown at this stage. However, previous analyses have shown that female's exhibit diminished knee joint proprioception in relation to males (27-30). Prophylactic knee sleeves, such as that used in the current investigation are proposed to promote stimulation of type δ sensory fibres within skin mechanoreceptors (31), and clinical research into their efficacy has shown that they are associated with improvements in knee joint proprioception (32-34). It can be speculated upon that there may be more scope for proprioceptive benefits in females, and that the positive effect of the knee brace in female athletes was mediated by a proprioceptive effect, which may have been responsible for the alterations in peak knee flexion that were evident only in female participants. Reductions in knee flexion are associated with lengthening of the moment arm of the patellar tendon itself, which leads to a reduction in tendon loading. Nonetheless, further mechanistic investigations into the specific effects of prophylactic knee sleeves on joint position sense at the knee are required before this notion can be recognized.

As stated previously, the aetiology of patellar tendinopathy in athletic populations, relates to the storage and release of energy by the tendon during sports movements (2). Therefore given the increased rate at which the tendon was loaded in the no-brace condition, this observation may have clinical significance. It can be conjectured that female athletes may be able to attenuate their risk from patellar tendinopathy during specific athletic movements through utilization of prophylactic knee bracing. However, further prospective research into the prophylactic effects of knee bracing is required before this can be clinically substantiated.

A further important observation from this investigation, is that for both male and female athletes, patellar tendon loading was significantly greater in the cut and hop movements in relation to the run condition. It is proposed that this observation relates to the ballistic nature of cut and single leg hop movements, in relation to the run condition, placing greater demands on the knee extensors. It has been shown through epidemiological analyses, that the aetiology of patellar tendinopathy is related to the magnitude of the loads experienced by the tendon itself (2). Importantly, cutting is one of the key abilities of sports games (35) and cutting actions are functionally specific to a range of different individual and team events including but not limited to; association football (36), American football (37), netball (4), tennis (38), squash (16) and basketball (39). In addition, single leg hop landings are similarly common in multidirectional sports including but not limited to; association football (40), American football (41), gymnastics (42), netball (7) and basketball (39). The findings from the current investigation indicate that cut and hop motions may place athletes at increased risk from patellar tendon pathology, therefore conservative prophylactic measures such as knee bracing may be important apparatuses in athletic disciplines and their associated training regimens whereby these movements are common. Future prospective research is clearly required to investigate the longitudinal prophylactic effects of different conservative modalities, in sports which place high mechanical demands on the patellar tendon.

A potential drawback to the current investigation is that patellar tendon loading parameters were quantified via a musculoskeletal driven model. Although this approach represents an advancement in relation to previous mechanisms, further progression is needed to improve the efficacy of musculoskeletal modeling of patellar tendon kinetics. Although muscle driven simulations of musculoskeletal loading require a range of mechanical assumptions, they have developed significantly in recent years. Thus, musculoskeletal simulations have the potential to become useful tools for clinical analyses in the field of biomechanics.

In conclusion, whilst previous analyses have investigated the therapeutic and prophylactic effects of knee bracing, the current knowledge with regards to the effects of prophylactic knee bracing on the patellar tendon in functional athletic movements is limited. The current investigation therefore addresses this, by examining the effects of wearing a prophylactic knee brace on patellar tendon loading parameters during run, cut and jump movements in male and female athletes. The current study showed firstly that patellar tendon loading parameters were significantly reduced in female athletes in the run and cut conditions whilst wearing the brace. In addition, for both males and females the cut and hop movements were associated with significantly greater tendon loading in relation to the run motion. Given the association between patellar tendon loading and the aetiology of patellar tendinopathy, this observation may be clinically important. It can be conjectured that female athletes may be able to attenuate their risk from tendinopathy during specific athletic movements through utilization of knee bracing, although further prospective research into the prophylactic effects of knee bracing is required before this can be clinically substantiated.

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Table 1: Patellar tendon load parameters (means, standard deviations and 95% confidence intervals) as a function of *brace* and *movement* conditions in male athletes.

	Male																	
	Run						***Cut***						***Hop***					
	Brace			*No-Brace*			*Brace*			*No-Brace*			*Brace*			*No-Brace*		
	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI
Peak patellar tendon load (BW)	7.03	1.25	6.24 - 7.83	7.48	1.48	6.54 - 8.42	8.08	2.03	6.80 - 9.37	8.30	1.46	7.37 - 9.22	7.76	1.67	6.69 - 8.82	8.07	1.22	7.30 - 8.85
Patellar tendon instantaneous load rate (BW/s)	335.41	115.57	261.98 - 408.84	358.54	114.05	286.07 - 431.01	445.64	162.25	342.55 - 548.73	457.89	153.72	360.22 - 555.56	442.39	184.86	324.94 - 559.85	518.55	270.58	346.63 - 690.49
Patellar tendon impulse (BW·s)	0.61	0.13	0.52 - 0.69	0.82	0.25	0.66 - 0.97	1.01	0.31	0.81 - 1.21	0.98	0.30	0.79 - 1.17	1.01	0.50	0.69 - 1.32	0.96	0.38	0.72 - 1.20

Table 2: Patellar tendon load parameters (means, standard deviations and 95% confidence intervals) as a function of brace and movement conditions in female athletes.

	Female																	
	Run						***Cut***						***Hop***					
	Brace			*No-Brace*			*Brace*			*No-Brace*			*Brace*			*No-Brace*		
	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI
Peak patellar tendon load (BW)	7.69	0.76	7.05 - 8.32	9.42	1.03	8.56 - 10.29	8.79	1.14	7.84 - 9.73	9.26	1.93	7.64 - 10.87	7.88	0.76	7.24 - 8.52	8.70	2.38	6.72 - 10.69
Patellar tendon instantaneous load rate (BW/s)	289.14	65.59	234.31 - 343.98	370.06	93.67	291.75 - 488.40	353.17	116.46	255.81 - 450.54	422.01	142.91	302.54 - 541.49	484.43	63.87	431.0. - 537.83	487.58	115.96	390.64 - 584.53
Patellar tendon impulse (BW·s)	0.79	0.10	0.70 - 0.87	1.00	0.07	0.94 - 1.05	0.95	0.12	0.89 - 1.05	1.05	0.19	0.90 - 1.25	0.84	0.09	0.76 - 0.91	0.99	0.42	0.64 - 1.34

Table 3: Knee flexion parameters (means, standard deviations and 95% confidence intervals) as a function of brace and movement conditions in male athletes.

	Male																	
	Run						***Cut***						***Hop***					
	Brace			*No-Brace*			*Brace*			*No-Brace*			*Brace*			*No-Brace*		
	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI
Angle at footstrike (˚)	10.92	4.34	8.16-16.68	13.30	5.98	9.50-17.10	10.26	4.48	7.42-13.11	12.67	5.76	9.01-16.32	12.94	6.29	8.95-16.94	13.70	3.16	11.70-15.71
Peak flexion (˚)	36.55	2.64	34.87-38.23	39.05	4.06	36.47-41.63	44.45	4.18	41.79-47.10	43.92	3.82	41.50-46.35	45.26	6.60	41.07-49.46	45.00	5.79	41.32-48.68

Table 4: Knee flexion parameters (means, standard deviations and 95% confidence intervals) as a function of brace and movement conditions in female athletes.

	Female																	
	Run						***Cut***						***Hop***					
	Brace			*No-Brace*			*Brace*			*No-Brace*			*Brace*			*No-Brace*		
	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI
Angle at footstrike (˚)	11.46	2.66	9.24-13.69	16.44	4.94	12.31-20.57	13.16	3.98	9.83-16.49	17.87	4.53	14.09-21.65	12.49	3.14	9.86-15.12	17.99	6.27	12.74-23.23
Peak flexion (˚)	36.64	1.92	35.04-38.25	41.12	3.84	37.91-44.33	44.35	2.12	42.85-46.12	45.71	3.12	43.10-48.32	49.74	8.48	42.65-56.83	53.39	11.50	43.78-63.00

List of figures

Figure 1: Patellar tendon forces as a function of brace and movement conditions – black = no-brace & grey = brace (a. = male run, b. = female run, c. = male cut, d. = female cut, e. = male hop and f. = female hop).