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Short communication

Patella tendon moment arm function considerations for human vastus lateralis force estimates

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ABSTRACT

In vivo muscle forces are typically estimated using literature-based or subject-specific moment arms (MAs) because it is not possible to measure in vivo muscle forces non-invasively. However, even subject-specific muscle-tendon MAs vary across contraction levels and are impossible to determine at high contraction levels without techniques that use ionized radiation. Therefore, different generic MA functions are often used to estimate in vivo muscle forces, which may alter force predictions and the shape of the muscle's force-length relationship. The aim of this study was to examine the influence of different literature-based patella tendon MA functions on the vastus lateralis (VL) force-angle relationship. Participants (n = 11) performed maximum voluntary isometric knee extension contractions at six knee flexion angles, ranging from 40° to 90°. To estimate in vivo VL muscle force, the peak knee extension torque at each joint angle was multiplied by the VL's physiological cross-sectional area (PCSA) relative to the quadriceps' PCSA (34%) and then divided by the angle-specific patella tendon MA for 19 different functions. Maximum VL force was significantly different across MA functions ($p \leq 0.039$) and occurred at different knee flexion angles. The shape of the VL force-angle relationship also differed significantly ($p < 0.01$) across MA functions. According to the maximum force generated by VL based on its literature-derived PSCA, only the VL force-angle relationships estimated using geometric imaging-based MA functions are feasible across the knee angles studied here. We therefore recommend that an average of these MA functions is calculated to estimate quadriceps muscle forces if subject-specific MAs cannot be determined.

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1. Introduction

Predictions of in vivo muscle force are crucial for improving our understanding of the mechanical costs of human locomotion (Patriarco et al., 1981), joint loading (Bergmann et al., 1993) and muscle function (de Brito Fontana et al., 2014). However, because it is not possible to measure in vivo muscle forces non-invasively, muscle forces are typically estimated from joint kinematics, surface electromyography (Lippold, 1952) and net joint torque or external force measurements. In these instances, the net joint torque measured around a joint is divided by an angle-specific generically-scaled or subject-specific muscle-tendon moment arm (MA) (de Brito Fontana and Herzog, 2016; Ichinose et al., 1997).

MAs are usually determined via one of three experimental methods. These methods include (1) the geometric imaging (GI) method (Herzog and Read, 1993; Smidt, 1973), (2) the direct load (DL) measurement method (Grood et al., 1984) and (3) the tendon excursion (TE) method (An et al., 1984; Buford et al., 1997). An additional method, based on anthropometrically-scaled subject-specific model calculations (MC), has also been reported (Arnold et al., 2010; Lu and O'Connor, 1996; Yamaguchi and Zajac, 1989). However, irrespective of the determination method, using literature-derived MAs has its limitations because MAs can vary between genders (Nisell, 1985) and are affected by the level of contraction (Arampatzis et al., 2004; Tsaopoulos et al., 2007), which means that subject-specific and generic MAs determined at rest or at low contraction levels might not actually lead to valid MAs at higher contraction levels.

A simple way to help legitimize the muscle forces estimated from generic literature-based or subject-specific MAs is to predict the maximum force a muscle is capable of generating based on its PCSA and specific tension. Ideally, PCSA estimates should be

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determined in vivo using ultrasound or MRI techniques on the participants of interest, or at the very least should be based on literature values from participants of similar age, gender and activity level as the study participants.

Due to the numerous MA functions in the literature and their specific limitations, the aims of this study were (1) to investigate how different literature-based generic MA functions affect VL force predictions and (2) the shape of the VL force-angle relationship, and (3) to examine which MA functions result in physiologically-appropriate maximum VL muscle forces, based on previously reported VL PCSAs and specific tensions of the quadriceps. We hypothesised that maximum VL muscle force estimates would vary across MA functions and occur at different knee joint angles, which would alter the shape of the VL force-angle relationship. We also hypothesized that not all maximum VL forces estimated using the literature-based MA functions would be physiologically realistic.

2. Methods

Eleven healthy male subjects (age 28.4 ± 2.6 years; height 182.1 ± 4.9 cm; body mass 80.8 ± 8.5 kg), free of ongoing and previous knee injuries, provided written informed consent prior to participating in the study. The experimental procedures were approved by the local Ethics Committee of the Faculty of Sport Science at Ruhr University Bochum and were in conformity with the Declaration of Helsinki.

Participants performed right-legged knee extension maximum voluntary contractions (MVCs) on a motorized dynamometer (IsoMed2000, D&R Ferstl, GmbH, Hemau, GER) whilst sitting in a reclined position with their hip fixed at 90° of flexion (similar to 85° in Erskine et al., 2009). The participant's lower leg was fixed with Velcro around the mid-shank to a cushioned attachment that was connected to the lever arm of the dynamometer (details shown in Fig. 1). Due to the knee joint rotation that occurs during knee extension contractions from rest to maximum effort (Arampatzis et al., 2004), the knee joint axis of rotation and dynamometer axis of rotation were aligned during MVC at 60° of knee flexion.

Participants attended one familiarization session where we ensured that the knee joint angles tested in the dynamometer set-up (40–90° in 10° increments) accurately reflected the actual knee joint angle during MVC as determined via an electronic goniometer (O'Brien et al., 2010), which consisted of a potentiometer (Vishay, Malvern, USA) with a linearity of ±2% and a precision of 1.5°. The goniometer's axis of rotation was aligned with the axis of rotation of the right knee, which was approximated by the lateral condyle, and the arms of the goniometer were aligned and fixed along the longitudinal axis of the tibia and femur with self-adhesive tape. The dynamometer crank arm angles that were used to attain the desired knee joint angles during MVC were then utilised in the following testing session. Participants performed three knee extension MVCs at the six knee joint positions outlined above. Knee extension MVCs were performed in a randomised order within a block that consisted of one MVC at each of the six knee joint angles. At least three minutes of rest was provided between contractions to minimise fatigue (de Brito Fontana and Herzog, 2016; Hahn et al., 2017). To account for the effects of gravity and passive joint torque on the net knee joint torque, five passive knee extensions ($5°s^{-1}$) were performed over the knee joint's range of motion while participants were instructed to relax. Torque and dynamometer crank arm angles were sampled at 1 kHz and synchronised using a 16-bit Power 1401 and Spike2 data collection system (Cambridge Electronic Design, UK).

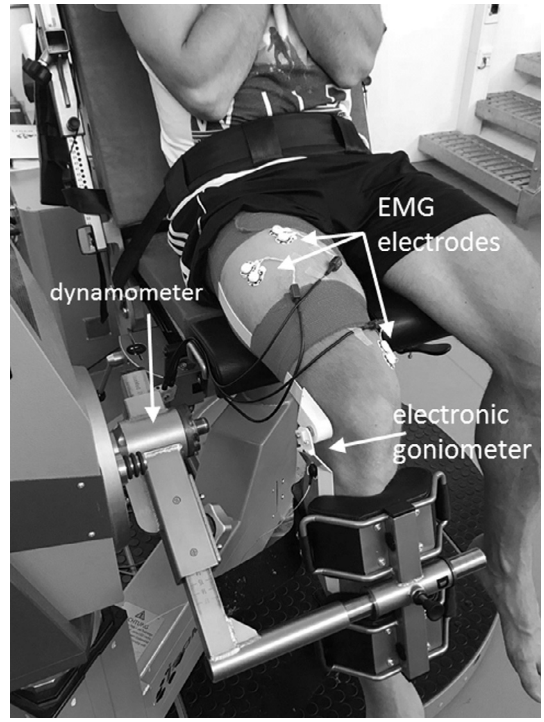

Fig. 1. Experimental setup with the participant sitting in the dynamometer at rest. The dynamometer was used to measure net knee joint torque and the lever arm crank angle. The dynamometer axis of rotation and knee axis of rotation (approximated by the lateral condyle) were aligned during maximum voluntary contraction. The electronic goniometer was aligned with the longitudinal axes of thigh and shank and used to measure the actual knee joint angle. EMG electrodes were used to measure VL, VM and RF muscle activity, but this data is not presented.

Raw torque and dynamometer and goniometer knee angle data were filtered using a dual-pass fourth-order 20 Hz low-pass Butterworth filter in Matlab (R2016b, Mathworks, Natick, MA). Filtered active net knee joint torque during the MVCs was calculated by subtracting the knee joint angle-specific filtered torque from the fifth passive knee extension (Esteki and Mansour, 1996; Lieber et al., 2000) from the peak torque value during the largest MVC from each joint position. According to the paper from de Brito Fontana and Herzog (2016), VL force (F_{VL}) was estimated as:

$$F_{VL} = \frac{M_{knee} * 0.34}{r_{knee}}$$

where M_{knee} is the maximum isometric active net knee extension torque, 0.34 represents the literature-derived VL PCSA relative to the quadriceps' PCSA (Akima et al., 1995), and r_{knee} is the knee joint angle-specific patellar tendon MA. Maximum active VL force for each tested knee joint angle was subsequently estimated using nineteen different literature-based patellar tendon MA functions, which covered the tested range of motion and included common methods to determine the patellar tendon MA (Table 1, Fig. 2). All VL forces estimated with the MA functions from the same determination method (i.e. GI, TE, DL and MC) were then averaged for each knee joint position to obtain the mean VL force for each determination method.

All data was tested for normality using Shapiro-Wilk normality tests. One-way repeated-measures ANOVAs and post-hoc comparisons with Bonferroni adjustments were performed to identify significant differences in estimated maximum VL force across MA functions. Two-way repeated measures ANOVAs were used to examine (1) differences in maximum VL forces at each knee angle across MA functions within a determination method

Table 1
Calculated maximum VL forces (mean ± SD) and knee joint flexion angles where maximum VL force occurred (angle (count)) for all moment arm functions compared in this paper and their method of moment arm determination.

MA function	*n*	Limb set-up	Method	Max. VL force	Knee angle VL Fmax
Smidt (1973)	26	In vivo	Geometric imaging	2312 ± 327	50 (4); 60 (6); 70 (1)
Grood et al. (1984)	5	In vitro	Direct load	4381 ± 690	60 (2); 70 (2); 80 (1); 90 (1)
Wendt and Johnson (1985)	10	In vitro	Direct load	2167 ± 326	50 (1); 60 (7); 70 (2); 80 (1)
Nisell (1985)	10	In vivo	Geometric imaging	2383 ± 335	50 (4); 60 (5); 70 (1); 80 (1)
Yamaguchi and Zajac (1989)	–	–	Model calculations	2646 ± 374	50 (4); 60 (6); 70 (1)
Marshall et al. (1990)	6	In vivo	Geometric imaging	2379 ± 344	50 (4); 60 (6); 70 (1)
Herzog and Read (1993)	5	In vitro	Geometric imaging	2287 ± 326	50 (4); 60 (6); 70 (1)
Baltzopoulos (1995)	5	In vivo	Geometric imaging	2802 ± 394	50 (4); 60 (6); 70 (1)
Buford et al. (1997)	15	In vitro	Tendon excursion	3609 ± 526	50 (4); 60 (6); 70 (1)
Chow et al. (1999)	1	In vivo	Geometric imaging	2576 ± 368	50 (4); 60 (6); 70 (1)
Kellis and Baltzopoulos (1999)	10	In vivo	Geometric imaging	2589 ± 373	50 (4); 60 (6); 70 (1)
Krevolin et al. (2004)	6	In vitro	Direct load	2449 ± 335	90 (11)
Kubo et al. (2006)	42	In vivo	Tendon excursion	2626 ± 422	60 (3); 70 (5); 80 (4)
Tsaopoulos et al. (2007) active	11	In vivo	Geometric imaging	2028 ± 291	50 (4); 60 (6); 70 (1)
Tsaopoulos et al. (2007) passive	11	In vivo	Geometric imaging	2158 ± 302	50 (4); 60 (6); 70 (1)
Pal et al. (2007) TE	–	–	Model calculations	3109 ± 435	50 (2), 60 (7); 70 (2);
Pal et al. (2007) GI	–	–	Model calculations	2293 ± 326	50 (4); 60 (6); 70 (1)
Arnold et al. (2010)	–	–	Model calculations	3111 ± 454	50 (1); 60 (6); 70 (2); 80 (1); 90 (1)
Son et al. (2018)	9	In Vitro	Tendon excursion	3122 ± 472	50 (3); 60 (6); 70 (2)

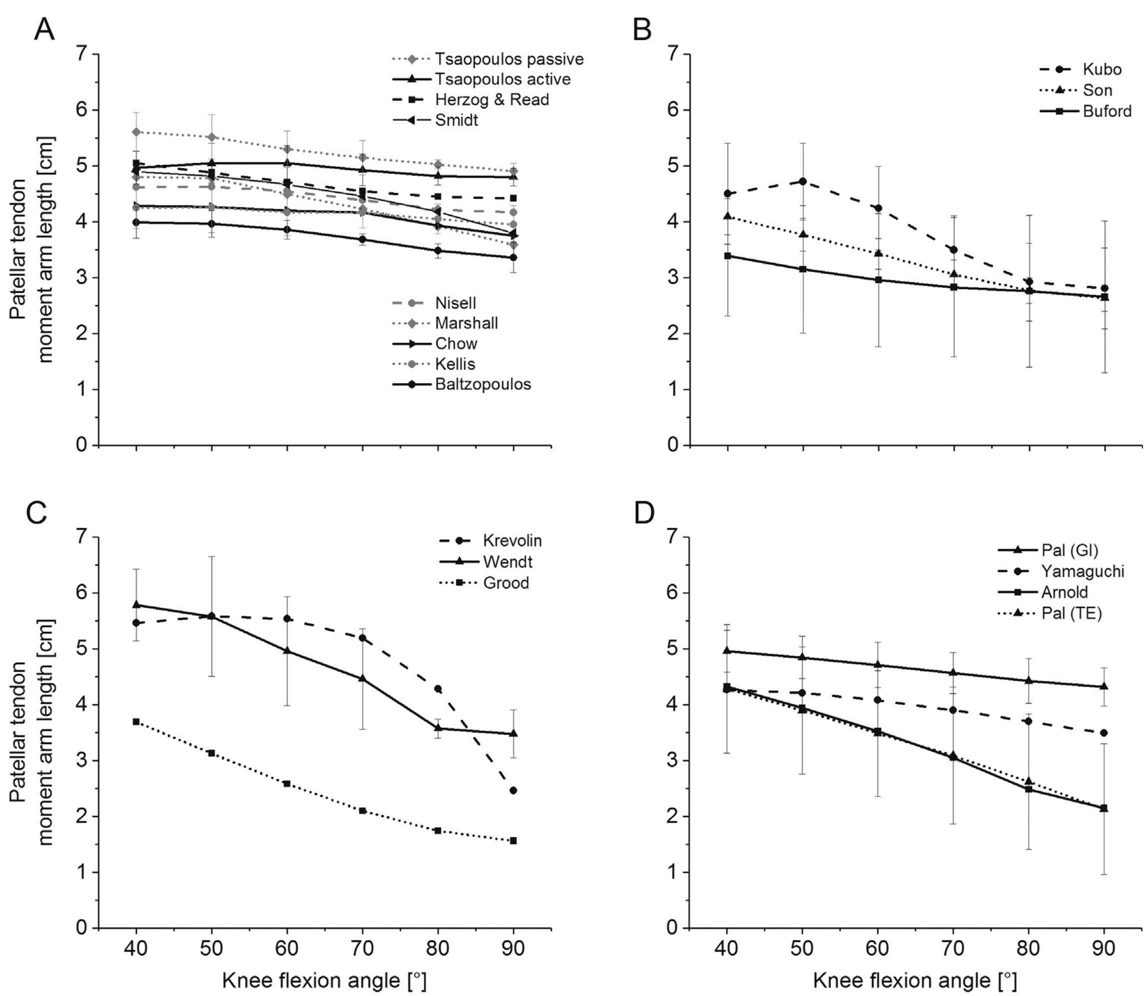

Fig. 2. Literature-derived mean patella tendon moment arm lengths separated into the four methods (A = geometric imaging method (GI), B = tendon excursion method (TE), C = direct load method, D = model calculations) that were compared in this paper across knee flexion angles from 40 to 90°. Data is presented as mean ± standard deviation (for those MA functions which provided standard deviations).

(within-method comparison; knee position × MA function) and (2) differences in the mean peak VL force at each knee angle across the averaged MA determination methods (between-method comparison; knee position × MA function). Significance was set at $p < 0.05$.

3. Results

Maximum active VL forces ranged from 2027.0 ± 291.7 N to 4386.7 ± 692.1 N (Table 1). Maximum VL forces across MA functions were significantly different (main factor: $p < 0.01$). After Bonferroni corrections, we found significant differences ($p \leq 0.039$) in maximum VL forces across MA functions in 137 of 171 comparisons. Maximum VL forces estimated by the MA functions of Buford et al. (1997) and Grood et al. (1984) were significantly different to all other MA functions.

The mean VL force-angle relationships determined by all MA functions within the GI, TC or MC methods resulted in an ascending limb from 40° to 60° knee flexion and a descending limb from 60° to 90° knee flexion (Fig. 3). However, the mean force-angle relationship determined from MA functions from the DL method did not show a clear descending limb (Fig. 3E).

A comparison of the VL force predictions for each tested knee joint angle across the MA functions from the same method revealed significant main effects of knee joint angle ($p < 0.01$) and MA function ($p < 0.01$), as well as a significant interaction ($p < 0.01$). The same comparison across the averaged MA functions between methods revealed significant main effects of knee joint angle ($p < 0.01$) and MA function ($p < 0.01$), and a significant interaction ($p < 0.01$). The shape of the VL force-angle relationship across MA functions within and between determination methods was therefore significantly different (Fig. 3A–E).

Knee joint rotation during the MVCs decreased from 12.9 ± 4.7° at 40° knee flexion to 3.6 ± 0.7° at 90° knee flexion (Table 2). Passive knee joint torque increased from −14.1 ± 2.2 N m at 40° knee flexion to 7.9 ± 2.1 N m at 90° knee flexion.

4. Discussion

The main purpose of this study was to investigate how different generic literature-based MA functions affect peak VL force predictions from 40° to 90° of knee flexion and the subsequent VL force-angle relationship obtained during MVCs. We found that different MA functions resulted in significantly different peak VL forces, which were not always physiologically-appropriate based on VL's literature-derived PCSA and quadriceps' specific tension. We also found significant differences in the shape of the VL force-angle relationships across MA functions within and between determination methods. These differences have serious implications for how we predict quadriceps muscle forces, which is important for estimating the mechanical costs of human locomotion and knee joint loads.

Our results show that the estimated peak VL forces were significantly affected by the selected MA function, as we found that peak VL forces significantly differed across MA functions in 81% of all comparisons. The differences in MA length (Fig. 2) significantly affected the VL force predictions across our tested knee joint angles and the variation in MA length between MA determination methods could be due to a combination of differing subject characteristics and experimental methodologies used to determine patellar tendon MA. We found that the averaged VL force-angle curves derived from the DL and MC methods were significantly different in shape to those derived from the GI and TE methods (Fig. 2E). This could be due to limitations of the determination methods. According to An et al. (1984), the DL measurement method assumes that friction loss and forces caused by surrounding soft tissues are negligible (An et al., 1984) and musculoskeletal models calculations (MC) often do not account for other rotational degrees of freedom apart from flexion/extension (Navacchia et al., 2017). Although the shapes of the averaged VL force-angle curves derived from the TE and GI methods were similar, force values were much higher from the TE method. The main assumption of the TE method is that the imposed joint angle rotation does not stretch the tendon (Olszewski et al., 2015). Baxter and Piazza (2018) found that Achilles tendon MAs determined from the GI method cannot be reproduced by simply scaling MAs estimated from the TE method and Fletcher and MacIntosh (2018) showed that the Achilles tendon MAs determined via the TE method were significantly larger when there was a passive moment that was unaccounted for.

Within the same MA determination method, we found that the VL force-angle relationships also differed significantly. This could be due to different ways of measuring MA using the same method, as Tsaopoulos et al. (2009) demonstrated for the GI method that implementing different knee joint centre of rotation definitions led to significant differences in patellar tendon MA lengths. However, low standard deviations in VL forces derived from the GI method across our tested knee joint angles (±130 to 227 N, Fig. 3E) and qualitatively similar shapes for all VL force-angle relationships derived from this method (Fig. 3A) show that choosing different GI-derived MA functions will not largely influence peak VL force predictions. Potential reasons regarding the larger deviations in VL force predictions within the other methods are, as summarized in Tsaopoulos et al. (2006), (1) anthropometric differences in the cadaveric or participant segment dimensions across studies; (2) differences in the condition of the tissues (in vivo vs. in vitro), and; (3) differences in the direction and magnitude of the external resistive forces and the internal ligament, muscle and other joint reaction forces.

In order to determine if our maximum VL forces estimated using literature-based generic MA functions were physiologically reasonable, we used data from Erskine et al. (2009), who found a mean VL PSCA of 75.1 ± 14.5 cm^2 and a quadriceps' muscle specific tension of 30.3 ± 4.9 $N\,cm^{-2}$, which multiplied together, results in a maximum force-generating capacity of ~2276 N for the VL. This is in accordance with other reports of VL's force-generating capacity, which ranges from 2052 to 2073 N (Ichinose et al., 1997; Son et al., 2018), and the mean VL PCSA is similar to the 74.0 cm^2 reported in O'Brien et al. (2010). By allowing for a ±10% difference in VL's estimated maximum isometric force capacity (2048–2504 N), which was selected based on a reported maximum standard deviation of 10.6% from GI-derived MAs as this method was performed on a larger sample size and had much smaller MA deviations than other methods (maximum deviation TE: 51%; DL: 57%; MC: 55%), we found that the mean maximum VL forces estimated from only 5 out of 9 MA functions that used the GI method (Herzog and Read, 1993; Marshall et al., 1990; Nisell, 1985; Smidt, 1973; Tsaopoulos et al., 2007) were physiologically reasonable. Similarly, the maximum VL force predictions derived from the MA function of Pal et al. (2007), who used model calculations based on the GI method, and those of Wendt and Johnson (1985) and Krevolin et al. (2004), who implemented the DL method, were physiologically appropriate.

We did not scale literature-based MA lengths to our participants' anthropometric characteristics because parameters to scale MAs cannot be easily quantified from surface measurements (Murray et al., 2002) and we did not want a scaling factor or model calculations to confound the interpretation of our results. Due to a greater mean height of our participants compared with the mean height of the participants we sourced our MA functions from, the mean VL forces in this study are likely to be underestimated and therefore our conclusions regarding which MA functions are

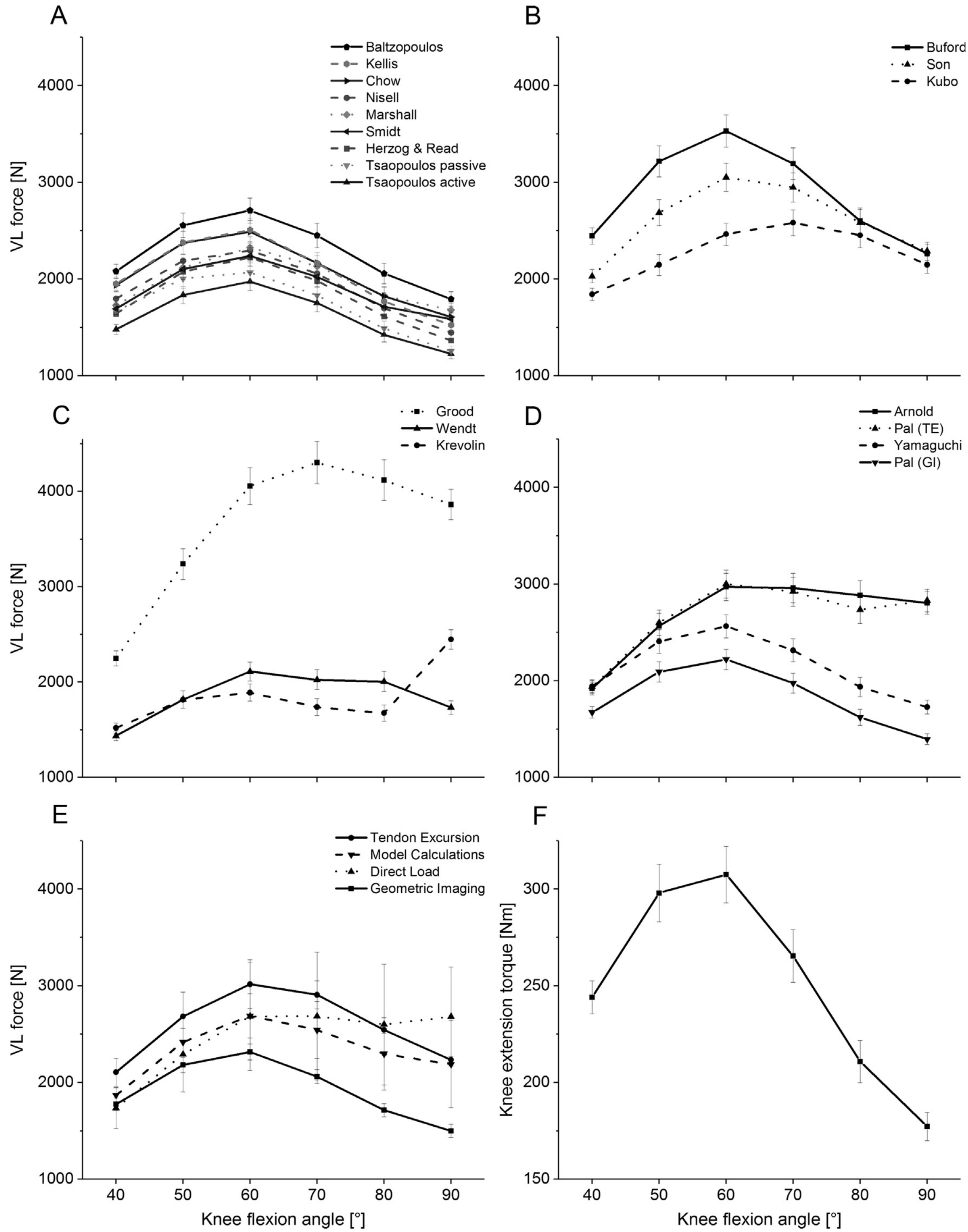

Fig. 3. Maximum active vastus lateralis (VL) forces across the tested knee joint angles estimated using various moment arm functions with different determination methods (A = geometric imaging method, B = tendon excursion method, C = direct load method, D = model calculations). (E) Maximum active VL forces across the tested knee joint angles averaged across all MA functions within the same determination method and (F) maximum active net knee extension torques across the tested knee joint angles. Data are presented as mean ± standard error.

physiologically appropriate are generally conservative. Although the approach we used to estimate VL muscle forces includes assumptions that limit them to being coarse estimates only, the differences we observed in VL force predictions were a direct result of the MA function, which was what our study sought to assess.

In conclusion, the main finding of this study was that different generic literature-based MA functions strongly affect maximum VL force predictions and its force-angle relationship under MVC. It appears that MA functions determined from the GI method result in the most physiologically-realistic in vivo VL force estimates based on the estimated maximum force-generating capacity of VL. Therefore, we recommend the use of an average of the MA functions derived from the GI method, where maximum VL force was within the physiologically-defined limits (Fig. 4 and Table 3).

Table 2
Knee joint rotation magnitudes (mean ± SD) during maximum voluntary contractions at the tested knee joint positions.

Knee joint position	Knee joint rotation
40°	12.9 ± 4.7°
50°	12.5 ± 3.6°
60°	9.7 ± 2.8°
70°	6.6 ± 2.2°
80°	5.0 ± 1.3°
90°	3.6 ± 0.7°

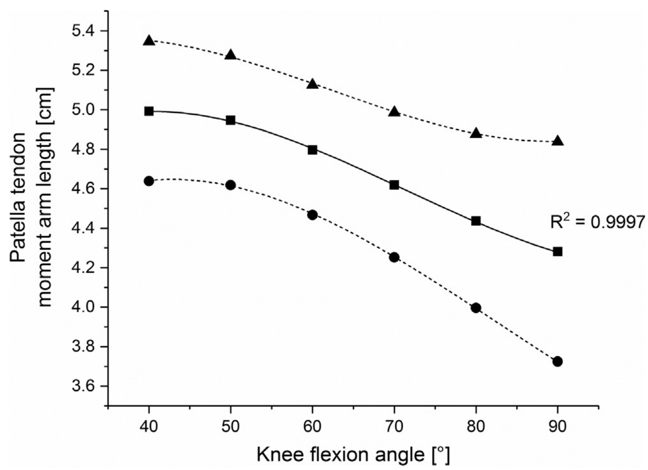

Fig. 4. Recommended moment arm (MA) function for the patellar tendon. Squares show the mean patellar tendon MA length across knee angles of five MA functions derived from the geometric imaging method, where maximum vastus lateralis (VL) force was within the limits of our estimated maximum VL force. Dots and triangles represent the lower and upper limits of the 95% confidence intervals (CI) of the recommended MA function. Third order polynomial regressions were fitted to the mean patellar tendon MA lengths (MA_L) (bold line; $MA_L = a^*KA^3 + b^*KA^2 + c^*KA + d$; $a = 6.852^*10^{-6}$, $b = 0.001465$, $c = 0.085$, $d = 3.494$) and upper and lower 95% CI (dashed lines) across knee angles (KA) from 40° to 90° of knee flexion.

Table 3
Mean patellar tendon moment arm (MA) lengths and their 95% confidence intervals (CI) at various knee angles for five geometric imaging derived MA functions, which had maximum vastus lateralis (VL) force estimates within the limits of our estimated maximum VL force-generating capacity.

Knee joint angle	Mean patellar tendon MA (cm)	Lower 95% CI (cm)	Upper 95% CI (cm)
40°	4.99	4.64	5.35
50°	4.95	4.62	5.27
60°	4.80	4.47	5.13
70°	4.62	4.25	4.99
80°	4.44	4.00	4.88
90°	4.28	3.72	4.84

Potentially, this MA function would also be appropriate for estimating muscle forces of other quadriceps muscles. Lastly, we recommend that participant characteristics (e.g. gender, age, anthropometric characteristics, activity levels) and loading conditions are comparable between the study or studies where MA is derived from and the study of interest. Future studies should investigate the potential of biplane fluoroscopy (Mozingo et al., 2018; Tsaopoulos et al., 2007) to quantify patellar tendon MAs at optimal knee angles for torque production at moderate and high contraction levels.

Conflict of interest

The authors have no competing interests to declare.

Author contributions

PB contributed to conception and design of the study, acquisition of data, data analysis and interpretation, and drafted the manuscript; BR contributed to conception and design of the study, data analysis and interpretation, and revising the manuscript critically; DH contributed to conception and design of the study, data interpretation, and revising the manuscript critically; All authors gave final approval for publication.

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