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Review

Human patellar tendon moment arm length: Measurement considerations and clinical implications for joint loading assessment

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Abstract

Detailed understanding of the knee joint loading requires the calculation of muscle and joint forces in different conditions. In these applications the patellar tendon moment arm length is essential for the accurate estimation of the tibiofemoral joint loading. In this article, different methods that have been used to determine the patellar tendon moment arm length under in vivo and in vitro conditions are reviewed. The limitations and advantages associated with each of the methods are evaluated together with their applications in the different loading conditions that the musculoskeletal system is subjected to. The three main measurement methods that this review considers are the geometric method, the tendon excursion method and the direct load method. A comparison of relevant quantitative results is presented to asses the impact of the errors of each method on the quantification of the patellar tendon moment arm and the implications for joint loading assessment in clinical applications.

Keywords: Moment arm length; Knee joint forces; Musculoskeletal modelling

1. Introduction

Accurate estimation of knee joint forces during various activities in competitive sport, exercise and rehabilitation is crucial for the prevention of injury or if the knee joint is to be protected and rehabilitated effectively after operative reconstruction. Although recent technological advancements in force and displacement transducers implantable to human tissue have allowed the measurement of the in vivo mechanical response of human knee joint structures (e.g. Finni et al., 1998, 2000), these techniques are highly invasive and hence impractical and not widely available. A more practical and non-invasive approach often used to estimate internal knee joint forces is based on mathematical models of the musculoskeletal system. In these applications, realistic representation of joint geometry and kinematics are of vital importance for accurate musculoskeletal loading quantification. An important biomechanical parameter that is determined by joint geometry and affects the muscle–joint system kinematics is the tendon moment arm length of the muscles crossing the relevant joint. During knee joint extension, the main moment arm affecting joint moment is that of the patellar tendon (PT). This is the leverage of the effective force transmitted to the tibia on contraction of the quadriceps muscle. The PT moment arm length is the perpendicular distance from the knee joint centre/axis of rotation to the PT action line and describes the tendency of the knee-extensor mechanism force to cause rotation of the tibia about the instant axis of rotation of the tibiofemoral joint (Pandy, 1999).

If the joint moment has been measured directly or calculated through inverse dynamics, then the moment arms and paths of the tendons determine the magnitude and orientation of the muscle–tendon forces relative to the bones or joint surfaces, for example, the orientation of the patellar tendon force relative to the tibial plateau. These are the main parameters used for the construction of relevant knee joint models and the calculation of the compressive and

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shear force components for the estimation of joint loading. The validity of the joint model and the accuracy of the calculated joint and ligament forces and, consequently, the validity of the joint loading conclusions will, therefore, depend largely on the precision and accuracy of the tendon moment arm length and orientation data. However, these joint kinematics parameters are difficult to measure directly on volunteers in vivo without the use of ionising radiation or other complicated imaging techniques. For these reasons, patellar tendon moment arm length and orientation data reported in the literature are normally used for the participants of different studies on knee joint loading. For example, in a study of tibiofemoral joint forces during open and close kinetic chain exercises, Wilk et al. (1996) used data reported by Herzog and Read (1993) and Spoor and van Leeuwen (1992) on muscle moment arm and orientation as a function of knee flexion angle to calculate compressive and shear forces. Zheng et al. (1998) developed an analytical model of the knee joint to estimate the knee forces during exercise. As part of this model, moment arms of muscle forces and angles of the line of action for muscles and ligaments were represented as polynomial functions of the knee flexion angle using data from Herzog and Read (1993). Schwameder et al. (1999) calculated the knee joint loading during downhill walking with and without hiking poles. They developed a two-dimensional knee model that was based on the data of Yamaguchi and Zazac (1989) to calculate quadriceps tendon forces, patellofemoral compressive forces and tibiofemoral compressive and shear forces. Bressel (2001) compared the patellofemoral joint force during forward and reverse bicycle ergometer pedalling. The quadriceps tendon force was calculated from the measured joint moment by estimating the effective moment arm based on the planar knee model and data reported by Yamaguchi and Zazac (1989). Thambyah et al. (2005) estimated the tibiofemoral forces during walking using data for the lines of action of the patella ligament and biceps femoris and their moment arms as functions of knee flexion angle based on the regression equations reported by Herzog and Read (1993). It is evident not only from the studies presented above as an example, but also from all the other studies that use a similar approach, that the PT moment arm length and orientation are very important parameters for sport and clinical applications because they affect the estimation of the joint compressive and shear forces and the conclusions about joint loading.

Several in vivo and in vitro experimental approaches have been described in the literature for obtaining the human PT moment arm length, based on kinematic measurements of various anatomical landmarks in living subjects, cadavers, or geometric models of the knee joint. This paper will concentrate on reviewing the above approaches and critically present their limitations, advantages and applicability under in vivo conditions relevant to the loading conditions that the musculoskeletal system is subjected to physiologically. Based on these characteristics, suggestions for future studies that should address important issues of clinical and practical significance will then be made.

2. Methods for PT moment arm length quantification

Three main methods have been used for quantifying the human PT moment arm length: (i) the geometric imaging method, (ii) the tendon excursion method and (iii) the direct load measurement method.

2.1. The geometric imaging method

With the geometric method the PT moment arm length can be measured in two (2-D) or three (3-D) dimensions. Moment arm measurements in 2-D rely on identification of an origin or reference point for the rotation of the segments. Such reference points have been (a) the instant centre of rotation (ICR), (b) the tibiofemoral contact point (TFCP), and (c) the anterior and posterior cruciate ligament intersection point (IP). Moment arm measurements in 3-D require identification of an axis around which the tibia or femur rotate with respect to each other. In both 2-D and 3-D representations, the PT moment arm length is measured in a way consistent with its definition, i.e., by measuring the shortest distance, in 2-D or 3-D space respectively between the origin/axis and the PT action line.

2.1.1. Two-dimensional representation

2.1.1.1. Application of the geometric method using the ICR. The ICR position in the tibiofemoral joint over a given knee joint rotation can be calculated using the Reuleaux graphical analysis (Reuleaux, 1875) on images taken in vivo showing the actual position of the tibia and femur at the two end-of-range angles of the rotation examined in the knee extension–flexion plane of movement. In this analysis, the femur is typically considered as being the stationary segment and the tibia the rotating segment. Two markers on the moving segment at the two end-of-range angles must be tracked and two perpendicular lines must be drawn from the midpoint of the line that connects each set of markers in the two positions. The point where these perpendicular bisectors intersect is the ICR of the tibiofemoral joint for that given knee rotation (Fig. 1). The PT moment arm length can then be obtained as the distance between the ICR and a line drawn along the mid-longitudinal axis of the PT tendon on an image at a mid-range angle of the initial rotation in the Reuleaux analysis. The relevant anatomical scans required can be obtained with X-ray and magnetic resonance imaging (MRI). One advantage of X-ray over MRI scanning is that measurements can be taken not only under static, but also dynamic conditions, e.g., knee rotation over a given range, which may better approximate the musculoskeletal system geometry during a specific habitual function of interest, e.g., walking or running. MRI scanning, however, apart from the no ionizing radiation also has the non-trivial advantage over X-ray of allowing a clear delineation of the PT trajectory due to

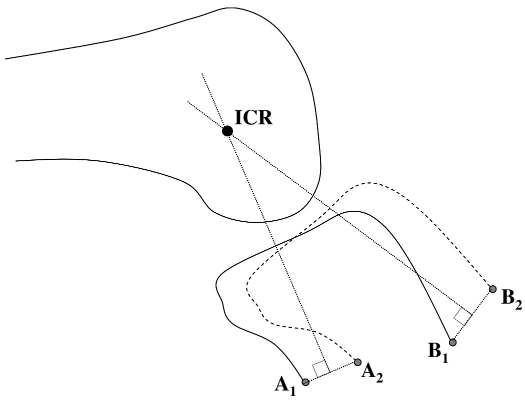

Fig. 1. Calculation of ICR in the tibiofemoral joint using the Reuleaux method. The intersection point of the perpendicular bisectors to the lines connecting points A_1 and A_2 and points B_1 and B_2 is the instant center of rotation of the tibiofemoral joint for the given knee rotation.

a marked tissue intensity contrast, although this necessitates long scanning durations in cases where the magnet strength is low. Clear soft tissue delineation cannot be achieved with conventional X-ray machines and an approximation of the PT path as a straight line between the tibial tuberosity and the patellar apex has, therefore, often been adopted. This approximation, however, cannot account for any potential buckling or curvature present along the tendon when slack, e.g., at knee angles corresponding to very short quadriceps muscle lengths at rest. A condition that needs to be satisfied in the application of the geometrical method using the ICR is that the imaging and joint movement planes coincide. A misalignment of the imaging plane with the knee rotation plane due to inappropriate relative placement of the knee joint in the X-ray or MRI scanner (2-D imaging) will alter the coordinates of the reference points and distort the view of the tibia and femur, allowing only their projection to the scanning plane to be seen. This will cause errors in the estimated location of the ICR (Panjabi and Goel, 1982; Spiegelman and Woo, 1987). A similar artifact will be caused if a non-planar movement of the rotating segment occurs during knee angular displacement. Calculation of the ICR in the tibiofemoral joint is liable to such artifacts because of the external rotation of the tibia round the femur during knee extension, the so called "screw-home mechanism". This phenomenon is due to the asymmetry between the two femoral condyles and it is profoundly apparent during the last 30° of open kinetic chain knee extension movements (Hollister et al., 1993). In close kinetic chain movements, the effect of the screw-home mechanism becomes negligible due to mechanical constrains imposed by the application of loading along the tibia (Levens et al., 1948; Lafortune et al., 1992; Koh et al., 1992), and the ICR location might therefore be less liable to the above errors. Moreover, a number of studies have shown that the accuracy of the ICR decreases as the rotation angle over which it is measured decreases (Panjabi and Goel, 1982; Woltring et al., 1985). According to Panjabi (1979) rotation intervals smaller than 10° can cause an error of about 9 mm in the location of the ICR. The PT moment arm relative to the ICR has been reported to be around 49 mm, in which case an error of the above magnitude can cause an underestimation or overestimation of 18% in the actual PT moment arm length. The determination of the locations of the ICR and PT action line is also affected by the accuracy level of the digitizing process. For example, using a Roentgen stereophotogrammetry system (RSA) for the estimation of the PT moment arm length and PT orientation, the measurement error could be approximately 0.01–0.1% of the entire field of view (FOV), while the measurement error for a typical videofluoroscopy system ranged from 0.13% to 0.25% of the FOV (Baltzopoulos, 1995).

A major advantage of the PT moment arm length quantification with the geometric method using the ICR is the feasibility of applying this method in vivo not only at rest (Smidt, 1973), but also during the application of loading. The latter is particularly important and relevant since there is evidence that, in contrast to what is implicitly assumed in most musculoskeletal modelling studies, tendon moment arm lengths during loading may substantially differ from those at rest (Maganaris et al., 1998, 1999). Clearly, if the loading on relevant joint structures is to be assessed realistically, then the tendon moment arm lengths used in the analysis must be measured in conditions approximating as closely as possible the loading and joint configuration examined. The application of loading, however, also introduces a problem. Due to the mechanical loading of the knee and the contact between segments, there are tibiofemoral joint reaction forces that can be resolved in the compressive and shear directions. The application point of these joint reaction forces is considered to be the tibiofemoral contact point (TFCP). The implication of this for the measurement of PT moment arm is discussed below.

2.1.1.2. Application of the geometric method using the TFCP. The TFCP is not an actual physical point, but is used as a representation of the contact surfaces between femur and tibia for simplification purposes. It is defined as the midpoint of the shortest distance between the surface of the two femoral condyles and the surface of the tibia plateau (Nisell et al., 1986, 1989; Fig. 2), and its position does not coincide with the ICR. This results in the tibiofemoral joint reaction forces creating moments around the ICR, but neither these forces nor their resultant moments can be calculated from measurements of knee extension joint moment using inverse dynamics analysis. Although the moment arm length of the shear and compressive force vectors with respect to the ICR can be quantified, the magnitudes of these forces cannot be determined from the moment equilibrium equation because the magnitude of the PT force is also unknown. This introduces additional unknown parameters in the moment equilibrium equation around the ICR, thus making it indeterminate. To circumvent this problem, several authors have applied the geometric method to calculate the PT moment arm length with

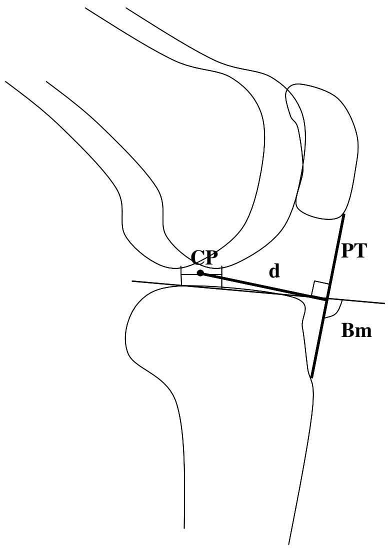

Fig. 2. Calculation of the PT moment arm length using the tibiofemoral "contact point". TFCP is the tibiofemoral contact point, d is the PT moment arm length and Bm is the angle between PT and tibial plateau.

respect to the TFCP (Lindahl and Movin, 1967; Nisell et al., 1986; Yamaguchi and Zazac, 1989; Herzog and Read, 1993; Baltzopoulos, 1995; Wretenberg et al., 1996; Kellis and Baltzopoulos, 1999; Lu and O'Connor, 1996). This approach makes feasible the solution of the moment equilibrium equation during knee joint loading because the moments of the unknown joint reaction forces relative to the TFCP are zero (Fig. 3). Furthermore, in contrast to identifying the ICR, locating the TFCP requires one image only, at the knee joint angle where the loading is applied. This eliminates several potential errors introduced by inappropriate selection of rotations and reference landmarks on the rotationary segment in the Reuleaux analysis (Panjabi and Goel, 1982; Spiegelman and Woo, 1987), reduces the imaging time, and makes the analysis simpler and faster. However, as emphasized above, it should be remembered that the TFCP is not the true ICR, but it is an easily identified reference landmark for the determination of the PT moment arm length and further studies are needed to quantify the errors introduced by this simplification.

Similar to the ICR, the location of the TFCP can also be calculated from X-ray and MRI scans. In addition, it can be located in cadaveric knees through direct digitization of relevant structures (Herzog and Read, 1993), or in knee models constructed from cadaveric measurements or scans of living humans at different knee joint angles (Yamaguchi and Zazac, 1989; Lu and O'Connor, 1996). This allows estimation of the moment arm over the knee joint rotation of interest as the shortest distance between the TFCP and the PT tendon action line.

2.1.1.3. Application of the geometric method using the IP. In some studies where the knee joint geometry has been modelled as described above, the PT moment arm length has been measured with respect to the IP. This preference is based on the readily discernible anatomical position of the IP usually based on a four bar-linkage model (formed by the tibial plateau, femoral surface, anterior cruciate ligament and posterior cruciate ligament) (Fig. 4), indicating that tibiofemoral relative movements around the IP approximate closely those around the true ICR. However, as is the case with the ICR, obtaining the PT force by solving the moment equilibrium equation around the tibiofemoral joint becomes impossible due to the introduction of the unknown moments of the shear and compressive contact forces relative to the IP.

One major advantage of mathematical models describing the dynamical mechanical behavior of knee joint struc-

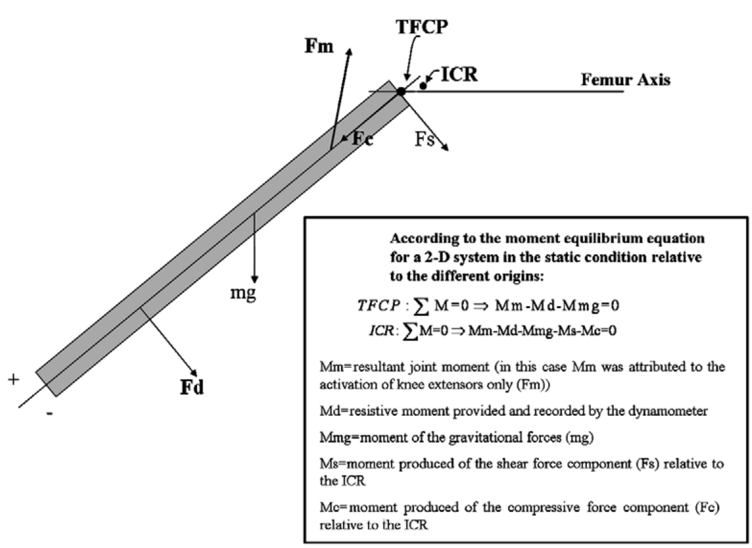

Fig. 3. Free body diagram of the tibiofemoral forces during knee joint loading (isokinetic knee extension). Using the TFCP the moment M_s and M_c of the unknown joint contact shear (F_s) and compressive (F_c) reaction force components become zero because the moment arm length of the F_s and F_c with respect to the TFCP is zero. The only unknown in the moment equilibrium equation is the knee joint moment (M_m). Using the ICR the moment equilibrium equation includes 3 unknown (M_m, M_s, M_c) and the system becomes indeterminate with no unique solution.

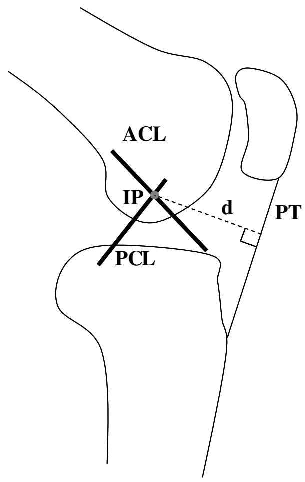

Fig. 4. Calculation of the PT moment arm length (d) using the anterior and posterior knee ligament intersection point (IP), where d is the PT moment arm length.

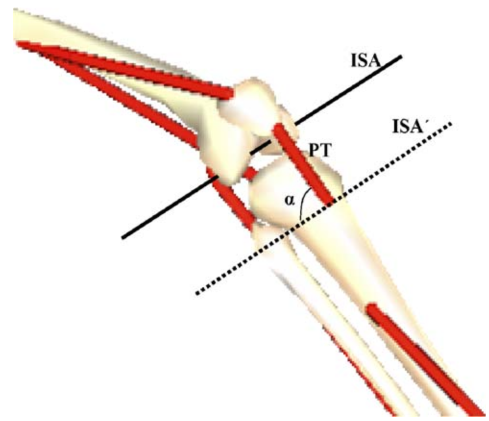

Fig. 5. Calculation of the PT moment arm length in three-dimensions according to Krevolin et al. (2004), showing that the magnitude of the PT moment arm length (PT_{ma}) is equal to the perpendicular distance between ISA of the knee joint and the PT action line (d) multiplied by the sine of the twist angle (α) ($PT_{ma} = d \times \sin\alpha$). ISA′ is a line parallel to the ISA.

tures, irrespective of whether they estimate PT moment arm lengths with respect to the TFCP or IP, is their applicability in simulating the effect of loading. This avoids the need for obtaining the co-ordinates of relevant anatomical structures during the actual loading via measurements on cadavers or medical images of the joint whilst being loaded. The general principle applying is that the more detailed the model, the more realistic the simulation of the knee joint. Model complexity, however, is inevitably associated with computational inefficiency and requires precise knowledge of the anatomy and mechanical properties and behaviour of all relevant structures. Due to these problems, practical, yet rather simplified models that fail to account for factors such as ligament and tendon elasticity (Gill and O'Connor, 1996), have often been presented. More recent models, however, have incorporated these properties and effects (Imran et al., 2000).

2.1.2. Three-dimensional representation

The presence of the screw-home mechanism during some activities (Hollister et al., 1993; Matsumoto et al., 2000; Nordin and Frankel, 2001), results in a relative tibiofemoral rotation not about a perpendicular axis to the PT action line, but around the instant screw axis of rotation (ISA), an axis that forms an angle smaller than 90° with the PT action line. This angle, the so called "twist angle", is greater than 60° for most of the knee joint range of movement. Its presence dictates that the PT moment arm length in 3-D is the perpendicular distance between the ISA and the PT action line multiplied by the sine of the twist angle (Fig. 5). Therefore, if for simplicity reasons the twist angle is considered to be 90°, as is the case with the 2-D methods described above, an overestimation proportional to the sine of the twist angle will be introduced in the calculation of the PT moment arm length. Despite the realistic orientation of the ISA, its use as reference axis to obtain the PT moment arm length during loading reintroduces again the problem of solving the moment equilibrium equation due to the unknown magnitude of the moments by the shear and compressive joint contact reaction force vectors.

Locating the ISA requires knowledge of 3-D co-ordinates of physical landmarks on the bony surface at different joint angles, which can be obtained with direct digitization on the relevant knee joint structures or indirect video-based image analysis. Although these approaches have so far been used on cadaveric specimens, obtaining 3-D coordinates of physical landmarks in vivo directly is also possible, e.g., by using as landmarks intracortical pins identifiable with biplanar (3-D) X-ray imaging or external cameras. Clearly, this approach is highly invasive. However, an indirect, noninvasive process based on model-fitting using 3D-to-2D image registration has recently been developed for single X-ray images and this allows 3-D tibiofemoral kinematics (e.g. Komistek et al., 2003). In this method, patient specific 3D knee models are developed from computed tomography data and using an optimisation algorithm the pose of the 3D models can be precisely matched to the 2D X-ray images at various flexion angles. This allows the estimation of the 3D kinematics from the 2D X-ray images.

2.2. The tendon excursion method (TE)

Using this method, the average knee-extensor mechanism moment arm length over a given knee joint rotation is calculated from the musculotendon excursion-to-knee joint rotation ratio, based on the principle of virtual work (An et al., 1984; Fig. 6). According to this principle: $F \times dx = M \times d\varphi$, where F is the force required to cause a tendon displacement dx by acting about a rotating joint, $d\varphi$ is the corresponding joint rotation produced, and M is the moment of the force F about the joint. From this equation it follows that M/F (=moment arm length) = $dx/d\varphi$. dx is the excursion of the whole musculotendon unit and therefore the TE method quantifies the average

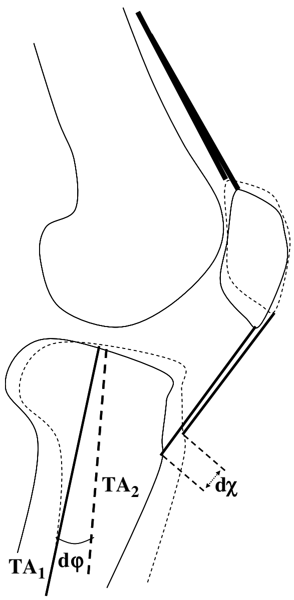

Fig. 6. Calculation of the knee-extenson mechanism moment arm length using the TE method. The initial and the final positions of the tibia are shown as solid and dotted outlines respectively. The angle between the longitudinal axis of the tibia (TA_1 and TA_2) at two different joint positions is $d\varphi$. The displacement dx represent the excursion of the musculotendon length over the knee rotation shown.

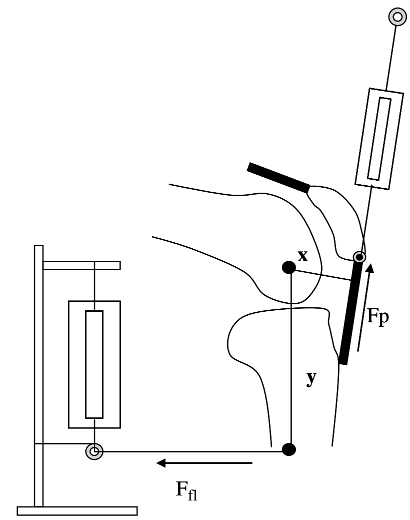

Fig. 7. Calculation of the PT moment arm length using the direct load measurement method. In the planar model an extension force F_p is applied to the PT through a spring scale. Flexion force F_{fl} is applied perpendicular to the longitudinal axis of the tibial. The magnitude of the F_p and F_{fl} can be read directly from the spring scales. In this system y is the flexion moment arm (must be measured directly on the cadaveric specimen) and x the PT moment arm length. Using the moment equilibrium equation the moment arm length can be written as $x = \frac{F_{fl} \times y}{F_p}$.

moment arm of the knee-extensor mechanism, but because dx is measured at the PT attachment level, it is usually considered as an approximation of the PT moment arm.

The TE method does not require knowledge of the position of the joint center of rotation or the tendon action line (An et al., 1984), but it requires planar joint movement, as do all 2-D methods, thus introducing all relevant errors discussed above relating to the consideration of 2-D kinematics as a realistic approximation of 3-D kinematics. Obtaining tendon displacements and using the TE method for obtaining moment arm length at the knee has only been possible until recently by direct measurement on rotating cadaveric limbs (Buford et al., 1997). Recent experiments (Maganaris, 2000) indicate that the application of scanning techniques such as ultrasound, MRI and X-ray may allow PT tendon excursion measurements under in vivo conditions. One disadvantage of this in vivo approach is that it can be applied only when there is joint rotation and therefore scans at more than one knee joint angle are required. In contrast the TFCP method can be applied on a single scan either at rest or during contraction. Another disadvantage of the approach is that it can be applied only when the muscles are inactive. This is because contraction results in tendon lengthening and storage of elastic energy. This effect is not accounted for by the principal of virtual work, thus prohibiting its application under active muscle conditions.

2.3. The direct load measurement method

Using the direct load measurement method, the PT moment arm can be calculated from the moment equilibrium equation, by dividing the resultant knee extension moment with the known force applied along the PT in a cadaveric limb (Fig. 7). This method has been applied for calculating the quadriceps tendon moment arm length (Kaufer, 1971; Grood et al., 1984), but no reports are available concerning the application of the method for calculating the PT moment arm length. An issue that is raised when applying the direct load method relates to the calculation of the moment of the external force. Knowledge of the external lever arm is necessary, which in turn necessitates identifying the knee joint centre of rotation. Kaufer (1971) assumed that this reference point coincided with the femoral attachment of the fibular collateral ligaments, but Grood et al. (1984) took into account the tibial plateau in their calculations. The direct method cannot be applied in vivo, unless implantable tendon transducers are used to quantify the force produced by isolated contraction of the quadriceps muscle (e.g. as a result of selective electrical stimulation).

3. Comparative results between studies

To assess the impact made by the errors of each methodological approach on the estimation of the PT moment arm length, a comparison of relevant quantitative results of the various studies that used different methodologies is necessary (Table 1).

The application of the geometric method with respect to the ICR has resulted in PT moment arm length values which range from a minimum of 38 mm to a maximum of 49 mm across the physiological range of knee joint movement (Smidt, 1973; Fig. 8). The corresponding values

Table 1
Maximum patella tendon moment arm length, the angular position where it occurs and method of calculation as reported in the literature (M: male, F: female)

	n	Age	Height (m)	Body mass (kg)		Peak moment arm (mm)	Knee flexion angle (deg)	Method
Smidt (1973)	26	28	1.76	82	In vivo	49	30	ICR
Krevolin et al. (2004)	6	–	–	–	In vitro	51.9	45	ISA
Imran et al. (2000)	–	–	–	–	2D-model	54	30	IP
Gill and O'Connor (1996)	–	–	–	–	2D-model	41.9	∼120	IP
Baltzopoulos (1995)	5	20.8 (SD 3. 9)	1.79 (SD 0.03)	79 (SD 7.2)	In vivo	39.87	45	TFCP
Kellis and Baltzopoulos (1999)	10	23 (SD 1.5)	1.74 (SD 0.04)	74 (SD 3.8)	In vivo	42.6	45	TFCP
Nisell et al. (1986) M	10	27	1.82	75	In vivo	46.2	60	TFCP
Nisell et al. (1986) F	10	23	1.67	59	In vivo	37.9	60	TFCP
Wretenberg et al. (1996) M	10	29 (SD 5)	1.81 (SD 0.06)	79 (SD 7.8)	In vivo	50.8	0	TFCP
Wretenberg et al. (1996) F	7	25 (SD 5)	1.65 (SD 0.03)	60 (SD 6.7)	In vivo	47.1	30	TFCP
Herzog and Read (1993)	5	79.2	–	–	In vitro	52.8	30	TFCP
Lindahl and Movin (1967)	15	–	–	–	In vivo	48	30	TFCP
Yamaguchi and Zazac (1989)	–	–	–	–	2D-model	43	40	TFCP
Lu and O'Connor (1996)	–	–	–	–	2D-model	47.9	40	TFCP
Buford et al. (1997)	15	55.9	–	–	In vitro	51.1	0	TE

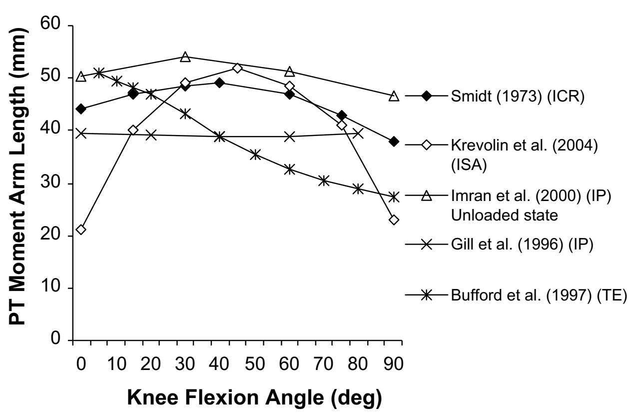

Fig. 8. Patellar tendon moment arm length as a function of knee flexion angle, estimated using the ICR, ISA, IP and TE methods.

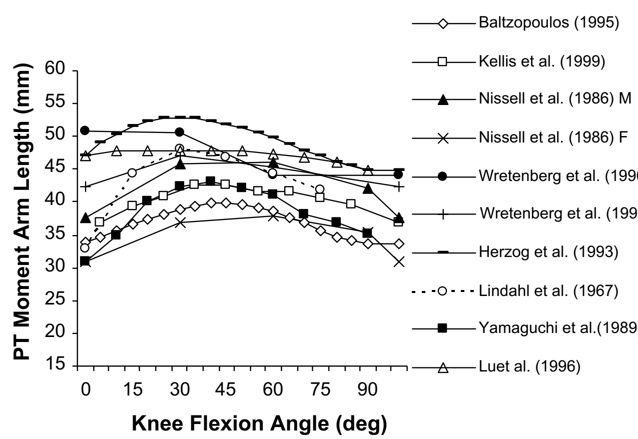

Fig. 9. Patellar tendon moment arm length as a function of knee flexion angle, estimated using the TFCP method (M: male, F: female).

resulting from the application of the geometric method with respect to the TFCP ranging from a minimum of 31 mm to a maximum of 52.8 mm (Lindahl and Movin, 1967; Nisell et al., 1986; Yamaguchi and Zazac, 1989; Herzog and Read, 1993; Baltzopoulos, 1995; Lu and O'Connor, 1996; Wretenberg et al., 1996; Kellis and Baltzopoulos, 1999; Fig. 9). Application of the IP method has resulted in average minimum and maximum values of 38.9 mm and 54 mm, respectively (Gill and O'Connor, 1996; Imran et al., 2000; Fig. 8). The TE method resulted in minimum and maximum values of ∼27 mm and 51 mm, respectively (Buford et al., 1997; Fig. 8). The 3-D representation has produced minimum and maximum values of ∼21 mm and 52 mm, respectively (Krevolin et al., 2004; Fig. 8). The variation in PT moment arm length with knee joint angle in the above geometric methods can be explained by (a) a change in the position of the origin/axis and (b) a shift in the path of the PT action line, during knee flexion–extension, which is caused by the relative movement of the tibia and femur, the patellar gliding on the femur and the operation of the screw-home mechanism.

Several studies have investigated the PT orientation at different knee joint angles, and a common finding is that the PT tendon becomes less oblique relative to the tibial plateau as the knee rotates from extension to flexion and it is acting at right angles relative to the tibial plateau near 90° of knee flexion (approximately 90°). (Nisell et al., 1986; Eijden et al., 1985; Yamaguchi and Zazac, 1989; Herzog and Read, 1993; Baltzopoulos, 1995; Lu and O'Connor, 1996; Gill and O'Connor, 1996; Kellis and Baltzopoulos, 1999; Imran et al., 2000; Fig. 10).

Differences in the absolute values of PT moment arm length at any given knee joint angle may be attributed to several factors, including (a) anthropometric differences in the participants or in the cadaveric segment dimensions in different studies, (b) differences in the state of the tissue (in vivo vs. in vitro), and (c) differences in the magnitude and direction of the external resistive forces and the internal muscle, ligament and other joint reaction forces. Using the results of the studies described above, the relation between the mean values of the peak moment arm and

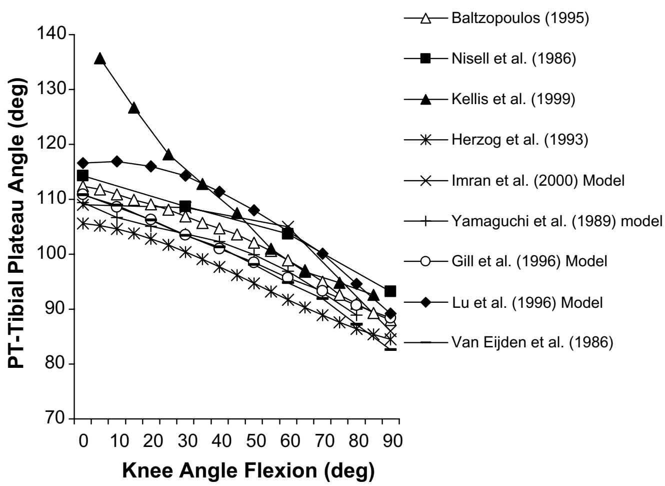

Fig. 10. PT-tibial plateau angle at a range of knee flexion angles from different studies.

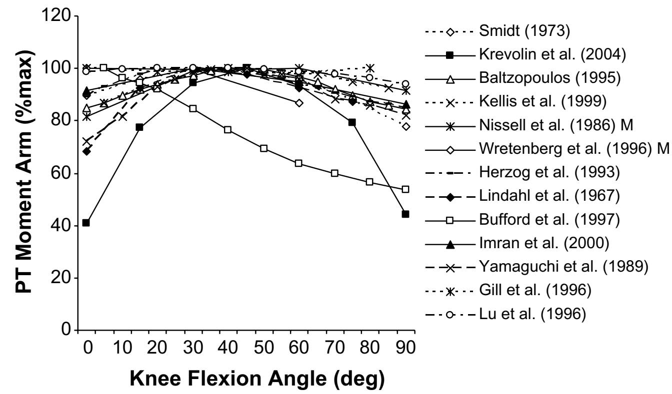

Fig. 12. Patellar tendon moment arm length from different studies normalised to the peak moment arm in each case (M: male).

the mean values of height and body mass of participants in the different studies is established and is shown in Fig. 11. Although a regression analysis cannot be performed because only the mean values of each parameter from the different studies are available, the relation that is presented in Fig. 11 and it is based on the mean values, indicates that the PT moment arm is likely to scale with height but not with body mass. Furthermore, open kinetic chain (OKC) knee extension as opposed to weight bearing knee extension produced significantly smaller posterior shear forces (Wilk et al., 1996) applied during the measurement. Hence, where joint loading assessment is required in clinical applications, it is important to use PT moment arm length values from previous studies that they have examined participants with similar anthropometric characteristics and under similar loading conditions. Otherwise the calculations may overestimate or underestimate the magnitude of the knee joint forces. Normalisation of each individual PT moment arm length data set to the respective maximum value obtained reveals a common pattern between most studies: the PT moment arm length is increased in the range 30–60° (0° is full knee extension) and is lower near full extension and flexion. The consistency is higher at mid-range angles, where there is a reduction of up to 10% in most studies, relative to the maximum value which is recorded in this range. However the variation between studies in the reduction relative to the maximum value is larger at extreme flexion and extension (Fig. 12). The reduction at the extremes positions of the range of motion (ROM) is up to about 20% for most studies, but there is also a reduction of 40–60% in some of the studies (Buford et al., 1997; Krevolin et al., 2004). The likely reasons for this difference are that Buford et al. calculated the PT moment arm using the TE method and Krevolin et al. (2004) in their study take into account the screw-home mechanism of the knee joint. The bell-shaped pattern of the PT moment arm length vs. knee joint angle relation partly justifies a similar pattern in the well established isometric knee extension moment vs. knee joint angle relation (e.g. Reeves et al., 2004).

4. Direction for future studies

It is evident from the studies reviewed above that the different imaging techniques, conditions and calculation methods used can produce different PT moment arm length measurements. The impact of using different methods on the estimation of the PT moment arm cannot be accurately assessed, unless all methods are applied on a given sample

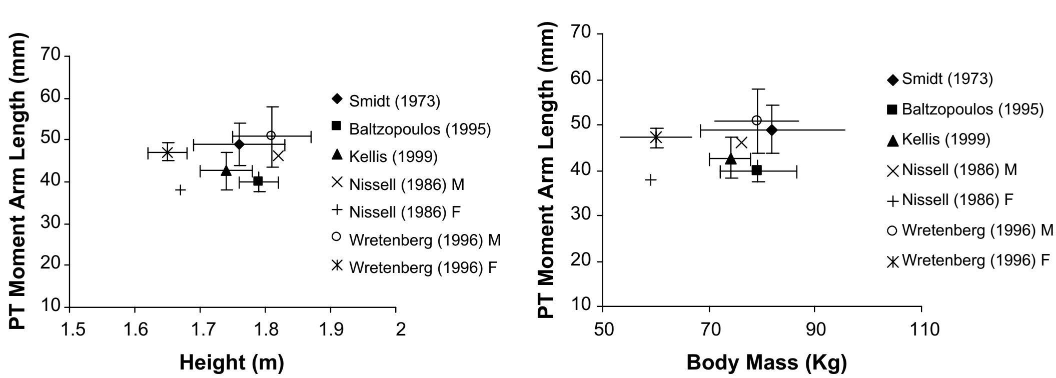

Fig. 11. Mean peak moment arm plotted as a function of the mean participants' height and body mass in each study (M: male, F: female).

to eliminate the potentially confounding effect of inter-subject PT moment arm length differences, e.g., differences due to body or joint size. Hence, a properly designed comparative study seems warranted.

So far, PT moment arm lengths have been obtained from cadaveric studies (Buford et al., 1997; Herzog and Read, 1993; Krevolin et al., 2004) or in vivo measurements at rest, with the knee joint unloaded (Smidt, 1973; Nisell et al., 1986; Wretenberg et al., 1996). In two in vivo studies the PT moment arm length was obtained during submaximal contraction of the knee extensors muscles, but neither the forces applied were quantified to ensure that they were constant across the range of movement tested, nor were the PT moment arm lengths obtained compared with rest to quantify potential differences between contraction states (Baltzopoulos, 1995; Kellis and Baltzopoulos, 1999). The moment arms of the human Achilles and tibialis anterior tendons during static maximal voluntary contraction increase by 22–44% compared with rest (i.e, relaxed muscles and unloaded joint), mainly due to changes in tendon orientation (Maganaris et al., 1998, 1999). Clearly, neglecting such sizeable contraction-effects will result in substantial errors in musculoskeletal forces estimated using modelling. For example, according to the data of Maganaris (2004) the calculation of force in the plantarflexor and dorsiflexor muscles from the moment equilibrium equation using resting moment arms overestimates the actual force by 22–44%. The corresponding error in the PT force calculation will be proportional to the change of the PT moment arm length from rest to the contraction state. Considering that in a simplified 2-D system the PT force is calculated by dividing the knee joint moment with the PT moment arm length, then error propagation theory suggests that the relative error of the calculated PT force will be the square root of the sum of the squares of the relative errors in the measurements of moment arm and moment. The magnitude of the error in the estimation of the compressive and shear knee joint forces depends not only on the change of the PT moment arm length but also on the change of the PT action line relative to the tibia from rest to the contraction state. The PT tendon moment arm length would be affected not only by changes in the PT orientation, induced for example by reducing the slack of the tendon by contraction, but also by changes in the position of the rotation centre/axis induced by the tibiofemoral contact forces. Since these forces depend non only on the magnitude of contractile force in the quadriceps muscle but also on the direction of the external load applied, studies on the effects of knee extension force applied under different mechanical external constrains (for example open- vs. close-kinetic chain knee extension) are required to assess the impact of these factors on knee joint mechanics in general, and the PT moment arm length in particular. Although the examination of increasing intensity static contractions would allow constructing a loading level-moment arm length relation from which the effect of submaximal dynamic contractions (for example those involved in walking) could be predicted, a direct measurement of moment arm length during the dynamic task of interest would be more relevant because this approach would account for differences in muscle activation and musculoskeletal loading in the time course of the task.

Estimation of the PT moment arm length in 3-D has so far been possible using cadavers. Similar principles could be applied to knee joint structures reconstructed from in vivo 3-D images, for example MRIs. A number of recent studies have indeed applied the above scanning approach and calculated 3-D moment arms in several human muscle–tendons (but not the PT) (Wilson et al., 1999; Graichen et al., 2001; Fowler et al., 2001). A major limitation with the use of 3D MRI is the long scanning time required, for example, durations of 4–10 min have been reported. Such durations, do not allow the continuous and constant application of contractile forces relevant to experimental and everyday life musculoskeletal loading, and as a result most MRI studies for PT moment arm measurements are performed at rest. The rapid progress in biomedical imaging technology will hopefully help circumvent these problems.

An issue of immense practical importance that has not be satisfactorily addressed so far in the literature is the prediction of the PT moment arm length from equations involving anthropometric parameters that can be easily measured without the need for sophisticated scanning procedures in each individual. A number of studies have attempted to address this issue. Visser et al. (1990) produced equations that describe changes of the moment arm length in several knee musculotendon units (but not PT specifically) as a function of limb length based on cadaveric measurements. Murray et al. (2002) employed in vivo scanning to predict changes in the moment arm length of elbow muscles as a function of the shorter distance between the elbow flexion axis and muscle origin and insertion. However measurement of this distance also requires scanning. Reid et al. (1987) presented a predictive equation for the moment arm length of the erector spinae muscle using in vivo scanning as a function of several anthropometric variables, including chest width, abdominal's least width and width at great trochanter. More recently, Krevolin et al. (2004) normalised the PT moment arm length by the femoral condyle width and the PT moment arm length–knee joint angle curves overlapped, indicating that bone size might be an appropriate scaling factor for PT moment arm length normalisation. Buford et al. (1997), however, found non-significant correlations between the PT moment arm length and several lower-limb bone dimensions including femur length, width and depth of the distal femoral condyle. Notwithstanding this inconsistency, that was probably due to the differences in the approach employed to estimate the PT moment arm length 3-D geometric method by Krevolin et al. (2004) and TE method by Buford et al. (1997), it is important to note that both of the above studies have been performed on cadaveric specimens. No attempt has so far been made to produce regression equations for the PT moment arm length as a function of relevant anthropometric variables based on surface measurements, such as limb length and related to joint size variables such as knee joint anterior–posterior or medio-lateral widths. Any such future attempt should consider not only the effects of contraction but also factors that can affect the mechanical qualities and the stabilizing effects of ligaments and other joint structures, since these will determine, to a large extent, the relative translation of the tibia and the femur and the shift in the location of the rotation axis (or any reference point used as rotation centre) in the transition from rest to the contracting state. Such factors include growth, aging, chronic physical activity and disuse (Butler et al., 1978; Vogel, 1991; Maganaris, 2004).

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