Achilles tendon moment arm length is smaller in children with cerebral palsy than in typically developing children

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Abstract

When studying muscle and whole-body function in children with cerebral palsy (CP), knowledge about both internal and external moment arms is essential since they determine the mechanical advantage of a muscle over an external force. Here we asked if Achilles tendon moment arm (MA_{AT}) length is different in children with CP and age-matched typically developing (TD) children, and if MA_{AT} can be predicted from anthropometric measurements. Sixteen children with CP (age: 10y 7m ±3y, 7 hemiplegia, 12 diplegia, GMFCS level: I (11) and II (8)) and twenty TD children (age: 10y 6mo ±3y) participated in this case-control study. MA_{AT} was calculated at 20° plantarflexion by differentiating calcaneus displacement with respect to ankle angle. Seven anthropometric variables were measured and related to MA_{AT}. We found normalized MA_{AT} to be 15% (~7mm) smaller in children with CP compared to TD children (p=0.003). MA_{AT} could be predicted by all anthropometric measurements with tibia length explaining 79% and 72% of variance in children with CP and TD children, respectively. Our findings have important implications for clinical decision making since MA_{AT} influences the mechanical advantage about the ankle, which contributes to movement function and is manipulated surgically.

Introduction

The mechanical advantage of a joint determines the internal muscle force needed to overcome an external resistance, e.g. ground reaction force. Mechanical advantage is defined as the ratio of internal to external moment arm (MA) length, and has been shown to influence joint and whole body function (Lee and Piazza, 2009). Differences in either MA can alter the outcome of muscle contraction, even if muscle function remains unaltered. In situations where movement function may be impaired and the study of muscle function is important, such as cerebral palsy (CP) (Barber et al., 2011), knowledge of MA length and mechanical advantage is vital if we are to understand the nature of impairment and how best to intervene.

Cerebral palsy commonly presents with bony deformities due to increased muscle forces acting on the bone (Morrell et al., 2002), which can cause 'lever-arm dysfunction' (Novacheck and Gage, 2007). At the ankle, an altered external lever-arm may be related to equinus, a midfoot break or tibial torsion, all of which shorten the external MA. Surgical interventions are recommended to increase external MA length (Gage et al., 2009). In the case of equinus, surgical lengthening of the Achilles tendon can improve kinematics, but weakness of the plantar flexors can occur as an adverse outcome of surgery (Gage, 2004). To understand the effects of surgical interventions that alter external MA, we need to consider how they will change the mechanical advantage. However, the length of the Achilles tendon moment arm (MA_{AT}), its contribution to mechanical advantage, and the influence it has on surgical outcomes is often not considered in clinical decision making.

The internal MA about which a muscle operates also determines more fundamental measures of joint function (Lieber and Friden, 2000). MA is proportional to maximum joint moment, typically referred to as muscle strength and the excursion range over which a muscle acts. Thus, differences in MA will affect the force-length properties of the muscle. Moreover, an altered muscle excursion over a similar range of motion (ROM) would also necessitate a different muscle shortening velocity, which influences the force-velocity characteristics (Lieber, 2002).

Furthermore, MA length is an important feature that determines the outcomes of musculoskeletal modelling. When using these models in children, typically developing (TD) or with CP, MA lengths are typically scaled down from adults using a 1:1 proportionality across all structures and dimensions (Sloot et al., 2015), but it is not clear if this is a valid way to scale (O'Brien et al., 2009; Waugh et al., 2011). Inappropriate scaling can lead to erroneous conclusions concerning muscle length and produced forces (Scheys et al., 2008).

In children with CP, the plantarflexor muscles are often affected by spasticity and contracture, and the foot and ankle joint by deformities. Currently, the focus lies on how these deformities affect the external MA, but we do not know how MA_{AT} may be altered. A previous study reported smaller muscle excursion over a common ROM in CP vs. TD children, which the authors hypothesised may be related to shorter MA_{AT} in the CP group (Matthiasdottir et al., 2014). However, since excursion was measured at the myotendinous junction (MTJ) and possible alterations in Achilles tendon lengthening during ankle joint rotation are unknown, the smaller muscle excursion may be caused by either a shorter MA_{AT} or greater Achilles tendon deformation. So far, no studies have directly measured MA_{AT} lengths in children with CP. Therefore, the purpose of this study was to quantify MA_{AT} length in children with CP and TD children using the tendon excursion (TE) method with tendon displacement measured at its distal attachment to the calcaneus. Also we wanted to establish if MA_{AT} in both groups can be predicted from anthropometric measurements. We hypothesized that MA_{AT} is smaller in children with CP compared with TD children, and that in both groups MA_{AT} is predictable from anthropometry.

Method

Participants:

Thirty-six children aged 6-16 years were recruited for this study. Sixteen children were diagnosed with spastic CP and twenty age-matched TD children served as a control group. Exclusion criteria were having botulinum toxin injection to the lower limb muscles within six months prior to testing or any lower limb orthopaedic surgery. All TD children were free from neuromuscular or skeletal disorders. Children were recruited through the gait lab of Alder Hey Children's Hospital in Liverpool and the University Hospital in Leuven. The study was approved by the Institutional as well as the NHS research ethics committee in the UK and the University Hospital's ethics committee in Leuven. The study was conducted in accordance with the Declaration of Helsinki. Written parental consent was obtained from the parents, and written assent was given by children in accordance with local regulations.

Procedure:

Participants lay prone on a bed with their leg in a custom made orthosis, to lock knee angle at 20° flexion and control ankle movement to occur only in the sagittal plane (Figure 1A). The axis of rotation of the orthosis was aligned with the lateral malleolus. The leg tested was the most affected, defined by spasticity scores, in the CP group and the left in the TD group. During each trial, the foot was passively rotated by an experimenter from maximum plantarflexion (PF) to maximum dorsiflexion (DF), taking five seconds to complete the ROM while ankle angle, calcaneus displacement, muscle activity and joint moment were measured.

This procedure was repeated six times with a minimum of 10 s of rest between each repetition. Participants were instructed to relax their muscles during the movements, which was checked for post-processing by inspection of surface electromyography (sEMG) of the Triceps Surae muscles at 1600 Hz (Zerowire, Cometa, Milan, IT). Forces and moments applied at the ankle were measured using a six degrees of freedom force sensor load-cell (ATI mini45: Industrial

Automation) attached to the orthosis under the ball of the foot. Ankle angle and calcaneus displacement along the length of the shank were calculated from the 3D position of two clusters of markers placed on the foot-plate of the orthosis and the shank and a single marker on the most superficial part of the posterior calcaneal tuberosity (Optitrack, US). Kinematic data was sampled at 120 HZ and kinetic data at 200 Hz.

Data processing:

Data analysis was carried out using custom made software (Matlab R2015a, Python 2.7.11). Kinematic and kinetic data were filtered using a 2nd order low pass Butterworth filter with a cut-off frequency of 6 Hz. Net ankle joint moment was calculated according to equation 1:

$$M_{joint} = -F_z d_z - F_y d_y - M_x - M_{orth}$$

Where F_z, F_y and M_x are the forces and moment exerted on the load cell in the z, y and x direction respectively, d_z and d_y are the moment arm distances from the point of force application of the load-cell to the lateral malleolus (see free body diagram, figure 1B) and M_{orth} is the predicted moment caused by gravity on the orthotic (Bar-On et al., 2013; Schless et al., 2015).

Calculation of MA_{AT}*:*

MA_{AT} was determined using the tendon excursion (TE) method (An et al., 1984; Ito et al., 2000), which defines MA as the ratio between linear displacement of the tendon and the change in joint angle ($MA_{AT} = \frac{\delta_{calc}}{\theta}$). Here, linear displacement of the Achilles tendon was defined at its insertion on the calcaneus, and measured from the displacement of a marker on the calcaneus along the direction of the tibia (δ_{calc}), corrected for marker size and skin thickness. A separate pilot experiment has shown that the marker on the skin was a valid representation of the distal insertion of the Achilles tendon (Supplementary material).

For each participant, calcaneus displacement vs. ankle angle curves were constructed for six passive dorsiflexion movements. In some children, a mid-foot break or heel lift out of the orthosis was apparent towards maximum dorsiflexion, identifiable by plateauing of the displacement-angle curve. In these cases, curves were trimmed to a smaller ROM. The calculated joint moments were checked over this trimmed ROM. At angles common to most participants, passive joint moments were typically ~1 Nm (Table 1) and were considered to be sufficiently low to not violate the assumptions of the TE method and allow us to continue the calculation of MA_{AT} at all angles (Olszewski et al., 2015). Next, a second order polynomial was fitted through the six trimmed displacement-angle relationships ($0.97 < r^2 < 0.99$). A second order polynomial allowed for the possibility for MA_{AT} to change with ankle angle. These polynomials were differentiated to construct a MA_{AT}-ankle angle relationship for each participant. These individual MA_{AT}-angle graphs within the trimmed ROM are plotted in Figure 2A. As the ROM over which MA_{AT} could be calculated was different for each participant, we selected MA_{AT} at one common ankle angle that most participants could achieve (-20°, with negative angles expressing plantarflexed position) for further calculations. Valid comparison of MA_{AT} at a single angle relies on equal slopes of the MA_{AT}-angle relationship between subjects. To test this assumption, the average gradients of the MA_{AT}-angle relations were compared between groups. In the CP group, one participant was excluded because the gradient of the MA_{AT}-angle relationship was determined as an outlier, i.e. exceeding '$Q3 \pm 1.5 * (Q3 - Q1)$', with Q1 and Q3 being the 25^{th} and 75^{th} percentiles. Furthermore, this profile was judged to be physiologically implausible (4cm change over a couple degrees). After removing the outlier, within groups, all gradients were within 2.7 SD from the mean and considered similar within and between groups ($p=0.27$, 95% CI [-0.1, 0.37]) (Figure 2B). These lengths were then corrected for marker size (11 mm diameter) by subtracting 5.5 mm. Finally, the individual distance measured from the surface of the skin to the attachment of the Achilles tendon on the calcaneus was measured with B-mode ultrasound and subtracted from the MA_{AT}.

Anthropometric measurements:

Seven anthropometric variables (Table 2) were measured by a trained physiotherapist: height, body mass, leg length (LL), tibia length (TL), foot length (FL), distance between medial and lateral malleolus (MD) and age. These parameters were assessed for predictive power and normalization of MA_{AT}. Tibial torsion was measured while the participant was lying prone with the knee in 90° flexion as the angle between the line of the longitudinal axis of the thigh and a line perpendicular to the axis connecting the most prominent points of the medial and lateral malleolus.

Statistics:

All parameters were checked to be normally distributed using the Shapiro-Wilk test and by inspection of the q-q plots. To determine if MA_{AT} could be predicted from anthropometric measurements, a backwards stepwise multiple regression analysis was performed as well as linear regressions between MA_{AT} and each anthropometric measurement separately, of which Pearson r^2 -values were calculated. Based on the r^2 -values of significant relationships it was decided whether, and to which, anthropometric dimension MA_{AT} could be normalized. Absolute and normalized MA_{AT} were then compared between groups with a 2-sample independent t-test. The level of statistical significance was set at $p \leq 0.05$.

Results

Data of fifteen children with CP and twenty TD children was used for the final MA_{AT} calculations. No differences were found between the groups for any of the anthropometric measurements. Participant characteristics can be found in Table 3. In the current sample of children with CP, TT was between 0°<α<20° (Median:15°, IQR:10-20°), which is similar to the tibial torsion range considered as "typical" (Mudge et al., 2014).

Correlations between MA_{AT} at -20° and each anthropometric measure showed mostly strong positive relationships ($p<0.01$, $r^2 = 0.36$-0.81, figure 3). Based on the high combined correlation across both groups, it was decided most appropriate to scale MA_{AT} to TL.

When MA_{AT} was normalized to TL, children with CP had significantly smaller MA_{AT} than TD children (p=0.001, 95% CI [1.05, 3.96]). This difference was consistent over the ROM studied (-30° to -5°) and its magnitude ranged from 1.4 to 2.5 %TL across the ROM. Absolute MA_{AT} were not significantly different between groups (p=0.0544, 95% CI [-0.15, 14.91]).

Linear regressions were performed with all anthropometric variables as separate predictors (Table 4). MA_{AT} could be significantly predicted by all variables independently in both children with CP and TD children. Inclusion of more variables in the regression did not improve predictive power (single regressions' $r^2 \sim 0.7$, vs. multiple regression $r^2 \sim 0.65$).

Discussion

The aim of this study was to quantify MA_{AT} in children with CP and TD children, and to determine whether this can be predicted from anthropometrics. We found that MA_{AT} in children with CP is 15% (~7mm) smaller than in TD children throughout an ankle ROM of -30° to -5°. Also, it was shown that in both groups MA_{AT} scales with, and can be predicted using a range of anthropometric measurements.

The relationship between internal and external MA around the joint reflects the mechanical advantage. This is an important feature when trying to explain ankle joint function in pathological gait. Considering a simplified scenario when a child is standing stationary on the forefoot, the smaller MA_{AT} in children with CP, would be associated with a smaller mechanical advantage at the ankle if it was not for the external MA also being smaller, for example due to equinus or other external MA deformities. The combination may actually mean that mechanical advantage does not turn out to be very different to typical (Figure 4A&B). Orthopaedic treatment, typically aiming to correct external MA length by optimizing foot shape and foot progression angle, will increase the external MA and thereby reduce the mechanical advantage. This would happen regardless of MA_{AT} length, but our results indicate that the mechanical advantage would not be restored to typical as intended, but in fact be reduced to less than typical (Figure 4C). This could be one of the contributing factors to the observed weakness of the plantar flexor muscles after equinus correcting surgery (Gage, 2004; Orendurff et al., 2002). Further research should focus on the specific influence of orthopaedic surgeries correcting lever arm dysfunction on the mechanical advantage of the ankle. Also, studies should seek to determine MA_{AT} length during gait, and under loading conditions (Rasske et al., 2016).

A smaller MA_{AT} also brings about changes in the moment-angle relationship. Shorter MA_{AT} would generate smaller maximal joint moments even when the same maximal muscle force is generated. Also, with a shorter MA_{AT}, the same muscle excursion will cause a larger joint rotation. Consequently, a joint moment can be produced over a broader ROM (Figure 5A). As less muscle excursion during joint rotation corresponds to lower angular velocities, joint moment would decline less as joint velocity increases (Figure 5B). Consequently, the shorter MA_{AT} of children with CP has the effect of creating joint function similar to that associated with a muscle with long fascicles and small physiological cross-sectional area (Lieber and Friden, 2000).

Spasticity is a major problem in children with CP, generally defined as a 'velocity-dependent increase in muscle activity' (Lance, 1980). Assessments of spasticity rotate the joint at high velocities to elicit a response (Bar-On et al., 2013). Since a shorter MA leads to a smaller muscle lengthening velocity when rotating the joint, relating angular velocities to muscle stretch will likely underestimate the true sensitivity of the muscle (Bar-On et al., 2014).

The smaller MA_{AT} in children with CP might be due to alterations in bony architecture or the development of bony deformities. Muscle thickness could also play a role (Maganaris et al., 1998), since this would influence the trajectory of the tendon's action line and so its distance from the centre of joint rotation. Children with CP have a smaller muscle cross sectional area than TD children (Barber et al., 2011), so this could contribute to the smaller MA_{AT}. Future studies should aim to confirm the exact cause of the decreased MA_{AT} in children with CP.

When using musculoskeletal models to study movement function, MAs are central in calculating muscle forces and lengths. Here we confirm that conventional scaling using a 1:1 proportionality across all structures and dimensions is not valid for scaling between adults and children. In addition, we show that applying any scaling algorithm similarly in children with CP and TD children is invalid. Therefore, we provide predictive equations to calculate MA_{AT} at -20° in children aged 6-16 years. We found the profile of the MA-angle relationship to be not significantly different within and between groups, this means that a) comparing MA_{AT} at a single joint angle is valid for the ROM described, and b) we can combine the predicted MA_{AT} with the known gradient (-0.9 vs -1.03 mm/deg, for CP and TD respectively) to calculate MA_{AT} at angles between -30° to -5°. More research is needed to extend these relationships over a full ROM.

Similar results concerning the predictability of patellar tendon MA have been shown in TD children (O'Brien et al., 2009), where tibia length was shown to be one of the main predictors. On the other hand, Waugh et al. (2011) concluded that due to low r^2-values, MA_{AT} could not be predicted from anthropometrics in TD children aged 5-12 years. The contrast with the present findings may be explained by the larger age range used in this study and thus an increased power to detect correlations.

This study has a few limitations. The sample size is relatively small for a heterogeneous population like children with CP. However, the statistical power of our comparison of MA_{AT} length between groups was 0.88, which suggests the sample was adequate for our purpose (Chow et al., 2003). Different methods can be used to calculate MA_{AT} length, and each has some limitations. In this study we used the TE method to define MA_{AT}, which has been shown to correlate well with MRI based methods (Fath et al., 2010). To exclude deformation of the tendon as a possible source of error we measured tendon displacement at the distal end of the tendon instead of proximally at the MTJ (Maganaris et al., 2000). Even with this step, the TE method used relies on several assumptions. First, it is assumed that the tensile load applied to the muscle-tendon was constantly low and thus it does not cause any deformations (Olszewski et al., 2015). We confirmed passive net joint moments to be low (~0-1 Nm) across the ROM studied, therefore we believe that no confounding tissue deformations have been introduced. Second, the TE method assumes ankle rotation to occur within the sagittal plane (Maganaris, 2004). However, the tibio-talar joint axis is typically externally rotated by an angle of 15±5°, which leads to an overestimation of MA_{AT}. In children with CP, tibial torsion can cause this angle to deviate from typical values. In the current study, tibial torsion, as measured by the bimalleolar angle, was within typical range, reducing the likelihood that it would have confounded our comparisons (Mudge et al., 2014). Unfortunately, this method of assessing tibial torsion has low correlation to more accurate radiological measures (Lee et al., 2009). To individually correct MA_{AT} for tibial torsion, better imaging techniques are necessary.

To conclude, this study has shown for the first time that Achilles tendon moment arm (MA_{AT}) is smaller in children with CP compared to TD children. This has important implications in clinical decision making since MA_{AT} influences the mechanical advantage about the ankle, which contributes to movement function and is manipulated surgically. We also found that both in TD children and in children with CP, MA_{AT} can be predicted from anthropometric measurements, which allows realistic quantification for relevant modelling applications without access to medical imaging facilities.

Conflict of interest

The authors disclose that they have no conflicts of interest.

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Table 1*: minimal, maximal and average joint moments measured at each ankle angle in the common ROM in typically developing (TD) and cerebral palsy (CP) participants.*

	-30°		**-25°**		**-20°**		**-15°**		**-10°**		**-5°**	
	TD	**CP**	**TD**	**CP**	**TD**	**CP**	**TD**	**CP**	**TD**	**CP**	**TD**	**CP**
Min (Nm)	-1.3	-0.9	-1.4	-0.9	-1.2	-0.8	-0.9	-0.6	-0.7	-0.4	-0.5	-0.2
Max (Nm)	-0.7	-0.1	-0.6	-0.1	-0.3	0.2	0.2	1.0	1.0	2.4	2.3	4.9
Mean (Nm)	-0.9	-0.7	-0.9	-0.6	-0.7	-0.4	-0.3	0.1	0.1	0.8	0.7	2.3

Table 2: *definitions of anthropometric variables*

Measure	**Method**
Tibia length	Distance from the tibiofemoral joint space to lateral malleolus
Leg length	Distance from the anterior superior iliac spine to the medial malleolus
Foot length	Distance from lateral malleolus to the head of metatarsal two
Inter-malleolar distance	Distance from the medial to lateral malleolus measured with 3d motion capture.

Table 3*: Participant characteristics*

Participant characteristics	**CP (n=15)**	**TD (n=20)**
Age (years)	11.1 (3.05)	10.4 (3.4)
Male/female (n)	10/5	11/9
Height (cm)	140.5 (20.6)	141.5 (16.6)
Mass (kg)	34.6 (18)	37 (14)
GMFCS (I-IV) (n)	10 I, 5 II	n/a
Diagnosis (n)	9 Diplegia, 6 Hemiplegia	n/a
Modified Ashworth Score (n=7) and Average Modified Tardieu (n=7) *	MAS: 1 (n=1), 1.5 (n=6), 3 (n=1) Tardieu: 2 (n=5), 3 (n=2)	n/a

Data are mean (SD) unless otherwise stated. CP: cerebral palsy; TD: typically developing; GMFCS: gross motor functional classification scale; n/a: not applicable.
*Tardieu scores from children recruited at Alder Hey Children's Hospital in Liverpool. MAS from children recruited at University Hospital in Leuven. One participant unknown.

Table 4: *predictive equations to calculate MA_{AT} at -20°, r^2 and p values for all anthropometric variables are shown. TL: tibia length, LL: leg length, FL: foot length, MD: distance between medial and lateral malleoli.*

	Formula	**r^2**	**p**
CP			
TL (mm)	MA_{CP} = -26.1 + 0.20 * TL	0.79	<0.01
Height (cm)	MA_{CP} = -38.7 + 0.57 * Height	0.80	<0.01
Body mass (kg)	MA_{CP} = 20.4 + 0.60 * Body mass	0.66	<0.01
LL (cm)	MA_{CP} = -26.2 + 0.93 * LL	0.81	<0.01
FL (cm)	MA_{CP} = -57.3 + 7.23 * FL	0.79	<0.01
MD (mm)	MA_{CP} = -27.9 + 1.13 * MD	0.67	<0.01
Age (y)	MA_{CP} = 2.45 + 3.52 * age	0.70	<0.01
TD			
TL (mm)	MA_{TD} = -8.2 + 0.17 * TL	0.72	<0.01
Height (cm)	MA_{TD} = -13.1 + 0.44 * Height	0.65	<0.01
Body mass (kg)	MA_{TD} = 28.9 + 0.53 * Body mass	0.68	<0.01
LL (cm)	MA_{TD} = -7.8 + 0.78 * LL	0.71	<0.01
FL (cm)	MA_{TD} = -22.1 + 5.00 * FL	0.57	<0.01
MD (mm)	MA_{TD} = -7.2 + 0.90 * MD	0.36	<0.01
Age (y)	MA_{TD} = 27.1 + 2.04 * age	0.55	<0.01

Figure Legends

Figure 1: A) Experimental setup of the ankle in the orthosis. Two clusters of markers were used to calculate ankle angle and a marker on the calcaneus was used to calculate calcaneus displacement. The load cell was used to calculate joint moments. During the experiment the ankle was moved from plantar to dorsiflexion. **B)** Free body diagram of the foot and foot plate. d_y and d_z correspond to the moment arm distances from the point of force application, respectively F_y and F_z, of the load-cell to the lateral malleolus. M_x is the moment exerted on the load cell in the x direction. $M_{orthosis}$ is the calculated moment caused only by the weight of the orthosis. The joint moment is given by: $M_{joint} = -F_z d_z - F_y d_y - M_x - M_{orthosis}$

Figure 2: A: Absolute Achilles tendon moment arm (MA_{AT}) length vs ankle angle for typically developing children (TD) and children with cerebral palsy (CP). Six angles common to most participants at 5° intervals between -30° and -5° were identified where we could record calcaneus excursion accurately. The boxes above each joint angle indicate the number of children in each group for which MA_{AT} could be calculated at that specific angle. -20° was the angle achieved by the most children. B: Box and whisker plot of gradient of the MA_{AT} vs angle curves in TD children and children with CP. * represents one outlier that was removed from further analysis. The edges of the box correspond to the 25th and 75th percentiles. The whiskers correspond to 2.7σ and cover 99.3 percent of all data.

Figure 3: Correlations between anthropometric variables and Achilles tendon moment arms (MA_{AT}) in children with cerebral palsy (CP) and typically developing children (TD). TL: tibia length, LL: leg length, FL: foot length and MD: inter malleolar distance.

Figure 4: Schematic drawing of the foot with the ground reaction force (GRF) and muscle force (F_m) acting over the external moment arm (MA_{ext}) and Achilles tendon moment arm (MA_{AT}) respectively. a: TD child, b: a child with CP in equinus and c: a child with CP where

surgery has lengthened the MA_{ext} but not altered the MA_{AT}. For simplicity, GRF and F_m are shown to act at right angles to the horizontal. In a and b, the ratio between MA_{AT} and MA_{ext} reflects the mechanical advantage and will be close to typical in the child with CP. c: When the MA_{ext} is lengthened the mechanical advantage will be decreased.

Figure 5: Schematic moment-angle (a) and moment-angular velocity (b) relationships of two identical muscles with different moment arm lengths. Muscles of children with cerebral palsy (CP) have smaller moment arms than typically developing (TD) children, all other muscle properties are considered to be identical. TD muscle will produce a larger peak moment for a given muscle force, however the active range of force production is larger in CP joint. Similarly, peak joint moment is larger in TD muscles, however, with increasing velocity the decline in joint moment is less in CP muscles.

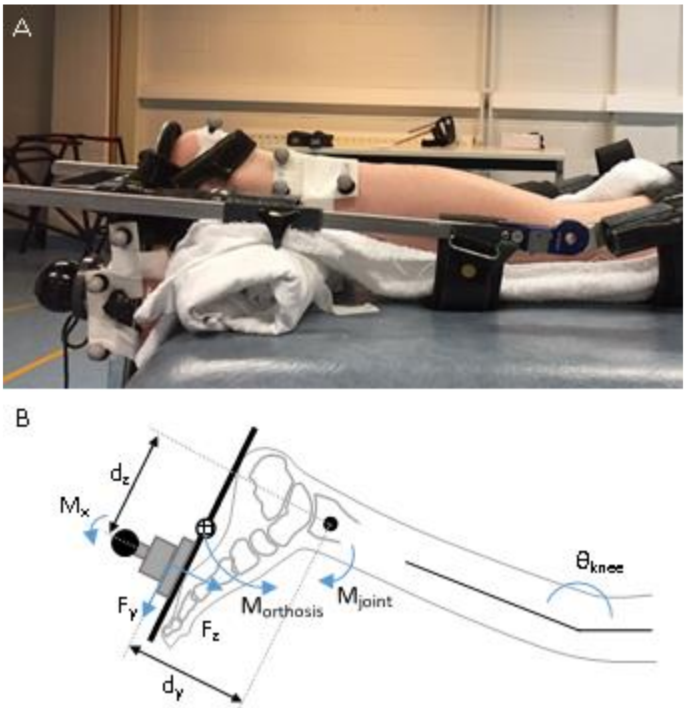
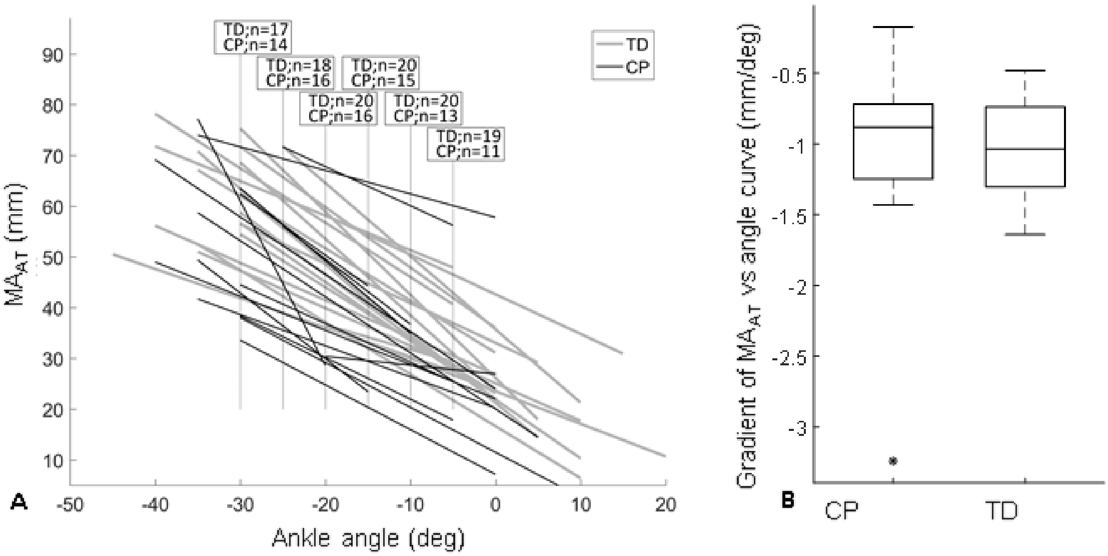


figure 1



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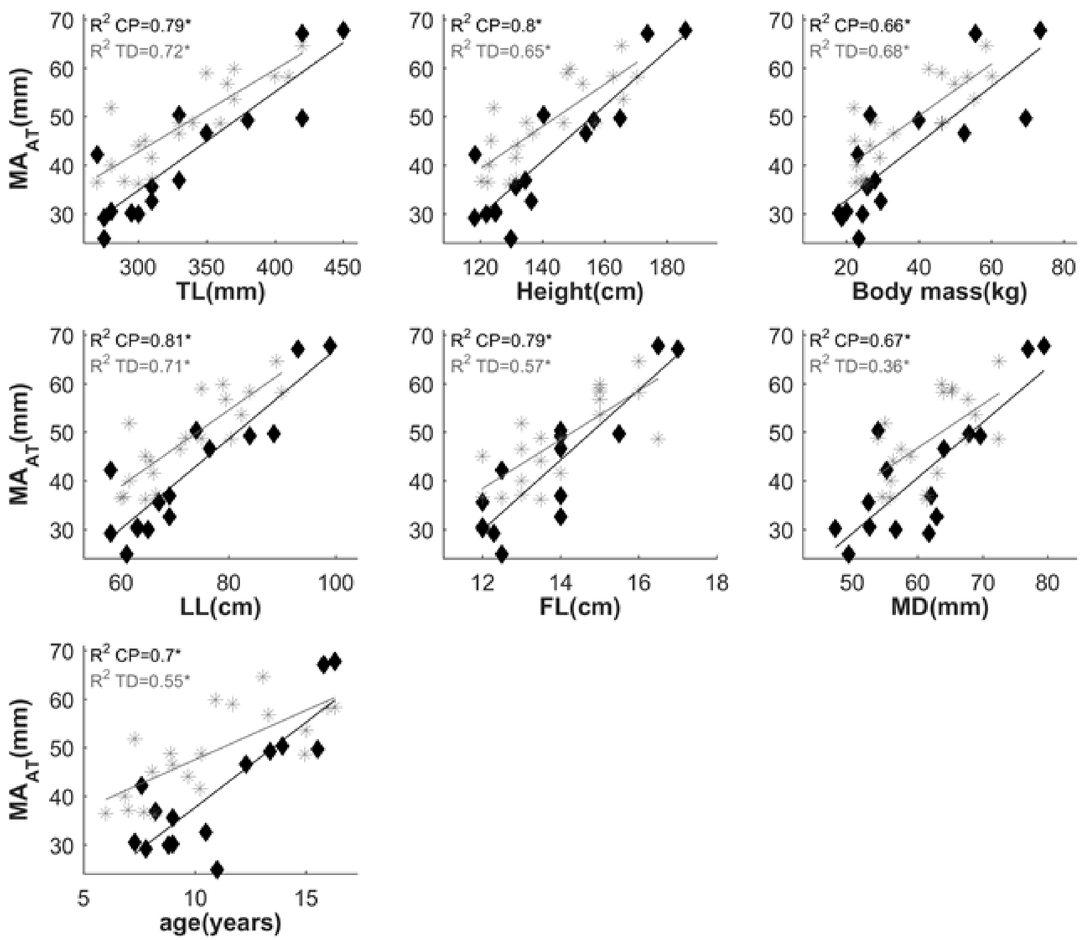

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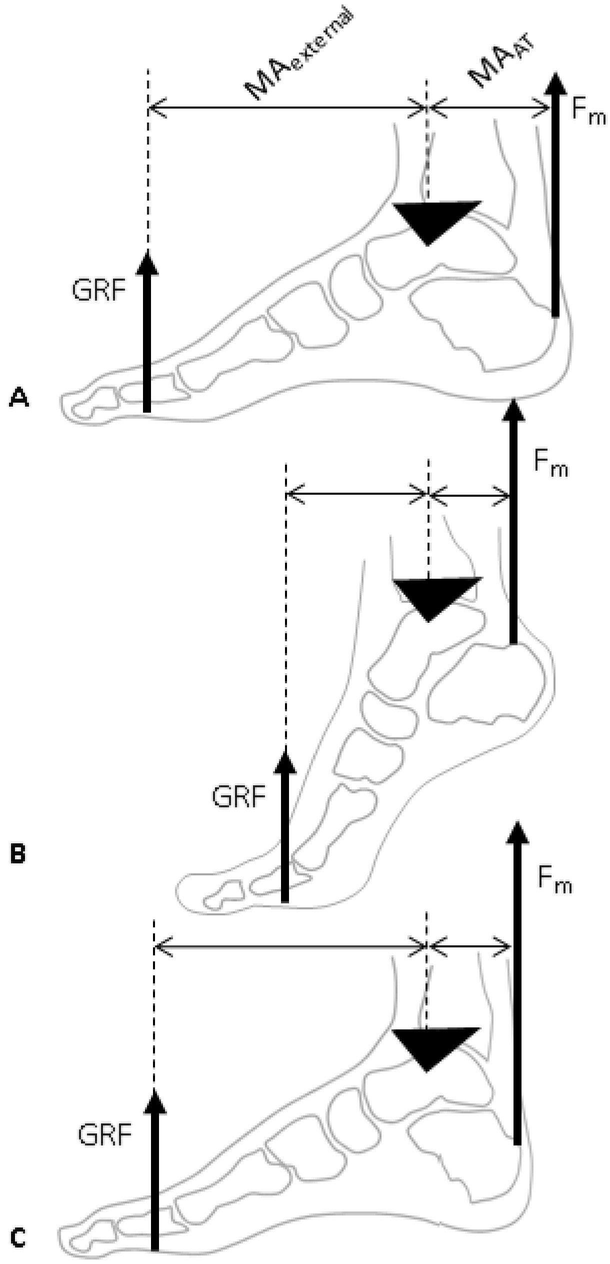

figure4

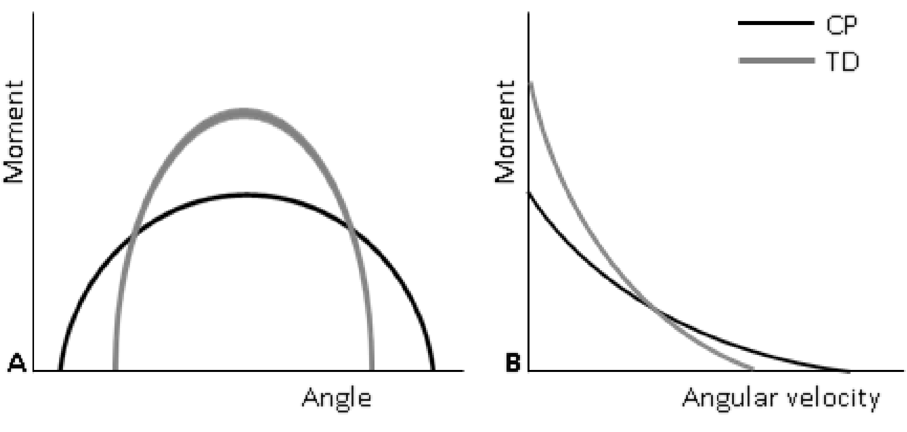

figure5