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Title: Prescribing joint co-ordinates during model preparation in OpenSim improves lower limb unplanned sidestepping kinematics

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Abstract (244 of 250 words)

Objectives: Investigate how prescribing participant-specific joint co-ordinates during model preparation influences the measurement agreement of inverse kinematic (IK) derived unplanned sidestepping (UnSS) lower limb kinematics in OpenSim in comparison to an established direct kinematic (DK) model.

Design: Parallel forms repeatability

Methods: The lower limb UnSS kinematics of 20 elite female athletes were calculated using: 1) an established DK model (criterion) and, 2) two IK models; one with (IK_{PC}) and one without (IK_0) participant-specific joint co-ordinates prescribed during the marker registration phase of model preparation in OpenSim. Time-varying kinematic analyses were performed using one dimensional (1D) statistical parametric mapping (α=0.05), where zero dimensional (0D) Root Mean Squared Error (RMSE) estimates were calculated and used as a surrogate effect size estimates.

Results: Statistical differences were observed between the IK_{PC} and DK derived kinematics as well as the IK_0 and DK derived kinematics. For the IK_{PC} and DK models, mean kinematic differences over stance for the three dimensional (3D) hip joint, 3D knee joint and ankle flexion/extension (F/E) degrees of freedom (DoF) were 46±40% (RMSE=5±5°), 56±31% (RMSE=7±4°) and 3% (RMSE=2°) respectively. For the IK_0 and DK models, mean kinematics differences over stance for the 3D hip joint, 3D knee joint and ankle F/E DoF were 70±53% (RMSE=14±11°), 46±48% (RMSE=8±7°) and 100% (RMSE=11°) respectively.

Conclusions: Prescribing participant-specific joint co-ordinates during model preparation improves the agreement of IK derived lower limb UnSS kinematics in OpenSim with an established DK model, as well as previously published *in-vivo* knee kinematic estimates.

Key Words: inverse kinematics; modelling; scaling; SPM;

Introduction to Conclusion (2,743 of 3,000 words)

Introduction

The sensitivity and specificity of a musculoskeletal model to quantify human movement is arguably one of the most important factors influencing the reliability of joint kinematics and kinetic estimates, as well as its ability to classify an individual's sport injury/re-injury risk[1]. In the field of biomechanics, there are two principal modelling approaches to estimate participant-specific kinematics: 1) direct kinematics (DK), and 2) inverse kinematics (IK). A DK modelling approach estimates frame-by-frame joint kinematics directly from markers placed on the skin of relevant anatomical landmarks, or digitized points held within technical coordinate systems[2]. An IK modelling approach estimates a model's generalized coordinates (i.e., joint angles) by fitting a participant-specific, rigid-body model to experimentally recorded kinematic data[3]. Most musculoskeletal models used in the field of biomechanics are variants of DK models developed approximately three decades ago[2], whereas the use of IK models within the biomechanics literature is comparably new[3]. The utility of IK models are thought to most beneficial to the field of sport biomechanics, where the forces and velocities of experimental movements are generally high, and the influence of soft tissue artefact (STA) on kinematic marker positions substantial[4-6].

An IK modelling approach addresses the issue of STA during high velocity sporting task by fitting a participant-specific, rigid-body model to experimental kinematics through segment or global optimization computational procedures[7]. With the release of the open-source musculoskeletal modelling software OpenSim[3], these complex, computationally cumbersome optimization processes are now user friendly, which is why they have gained large-scale exposure and uptake in the field of sport biomechanics and biomechanics as a whole. Since its release in 2007[3], recent estimates at the time of this publication have shown OpenSim has supported over 100,000 people worldwide. Though a positive step forward for the field of sport biomechanics, limited research has assessed the accuracy of IK derived kinematics relative to established DK models during high velocity, high injury/re-injury ACL risk sporting movements like single leg landing or unplanned sidestepping[8]. This places practical limitations for its use within the field of sport biomechanics as our ability to compare IK derived findings to previous research, share data between laboratories and replicate or externally validate research findings has arguably been unexplored.

Robinson et al.[9] directly compared DK and IK estimates of the knee during an unplanned sidestepping task (UnSS), with notable kinematic differences observed in the abduction-adduction degree of freedom. It has been purported in previously in the literature that model preparation or calibration can significantly influence IK derived upper limb kinematics[9]. As Robinson et al.[9] did not publish their model preparation methods, it is uncertain if their observed knee kinematic differences were due to the modelling approach used (DK vs. IK), their model preparation methods (prescribed vs. non-prescribed joint co-ordinates) or a combination of both.

There is considerable theoretical potential for IK modelling approaches to improve the reliability of lower limb kinematic estimates during high velocity sporting movements. Therein this research has the ability to facilitate the sport biomechanics fields' goals of standardised motion data modelling, data sharing and external validation procedures to substantiate experimental results. In addition, IK in OpenSim can provide an open, standardised platform for performing multi-centre clinical or prospective trials on a global scale. In an effort to help standardize IK modelling approaches in the field of sport biomechanics, the aim of this research was to determine whether prescribing participant-specific joint co-ordinates during model preparation in OpenSim influences IK derived UnSS kinematics. The rationale for using an UnSS task is because ACL injury events are through to occur during unplanned versus planned change of direction tasks[8-10]. No explicated hypotheses are presented as this is an exploratory time varying analysis.

Methods

Twenty elite female hockey players from the Australian Women's Hockey team (22±2.5 years, 1.7±0.1m, 63±6.3 kg, 22±4.3 BMI) attended one to four independent biomechanical data collection sessions, within a 13-month timeframe, at the University of Western Australia's (UWA) sports biomechanics laboratory. In each case, only an athlete's most recent testing session was used for analyses. All participants were injury free at the time of testing and provided informed consent in accordance with the UWA Human Research Ethics Board (RA/4/1/5713). This sample was one of convenience. An *a priori* power analysis could not be performed, as this is the first study in the field to perform a parallel forms repeatability analysis on time varying lower limb kinematic data.

Equipment, laboratory setup and experimental procedures were replicated across the four data collection sessions, as described by Donnelly et al.[11]. Motion capture data were recorded using a 20-camera hybrid Vicon MX/T40 system (Oxford Metrics, Oxford, UK) at 250 Hz, synchronized with a 1.2 m x 1.2 m AMTI force plate (Advanced Mechanical Technology Inc., Watertown, MA) sampling at 2,000 Hz.

Participants completed a random series of pre-planned and unplanned straight-run, crossover-cut and sidestepping tasks with their self-selected preferred stance limb. For all running and change of direction tasks, a trial was considered successful if their average approach velocity was between 3.5 ms^{-1} and 4.5 ms^{-1}. During the change of direction tasks a trial was considered successful if that changed direction at an angle 45±5° relative to their approach. These testing procedures have been shown to be repeatable between independent testing sessions[11].

Participants continued testing until five successful trials of each task were collected. The UnSS trials were then isolated, and used for further analysis.

Marker trajectories and ground reaction force (GRF) data were both low pass filtered at 14 Hz[12,13] using a zero-lag 4th order Butterworth in Vicon Nexus 1.8.1 (Vicon Peak, Oxford Metrics Ltd., UK). The choice of cut-off frequency was selected based on a residual analysis and visual inspection.

Ankle joint centers were defined using anatomical landmarks on the medial and lateral malleoli. A functional knee (mathematical helical) axis used to define the position of knee joint centers and the orientation of knee flex-ext axes[14]. A functional spherical method was used to define the position of hip joint centers[14]. A custom foot alignment rig facilitated the measurements of calcaneus inversion-eversion and foot abd-add to assist in defining the anatomical co-ordinate system of the bilateral shank and foot segments.

All static calibration and dynamic UnSS trials were processed using the aforementioned DK model, the lower limb kinematic and kinetic repeatability of which has been published previously[14]. All recorded marker trajectories and joint centers were exported into OpenSim 3.2.0 using a custom MATLAB software. An 8 segment, 20 degrees of freedom (DoF) rigid-linked skeletal model formed the foundation of the IK analyses in OpenSim v2.4[15]. Prior to IK modelling, the foundation model was prepared twice for each participant. These models, referred to as IK_0 and IK_{PC} for the purposes of this manuscript, were prepared as follows: 1) for the IK_0 model, segment lengths were scaled to participant-specific joint centers and marker registration was performed without participant-specific prescribed joint co-ordinates, and 2) for the IK_{PC} model, segment lengths were scaled to participant-specific joint center positions and marker registration was performed with participant-specific prescribed joint co-ordinates. During scaling, marker weightings were given a value of 1.0, with joint centres given a value of 1,000 for the IK_0 and IK_{pc} models. For the IK_{pc} generalised coordinates, a weighting of 100 was given.

Following model preparation, both the IK_{PC} and the IK_0 models were employed for kinematic analyses. Using the same UnSS trials processed using the DK model, IK was performed for the IK_0 and IK_{PC} models to obtain participant-specific lower limb joint angles. To allow for comparison to previous literature[9], weightings of 1.0 were used for all kinematic markers. To be clear, the kinematics from the DK model were only used during the model preparation for the IK_{pc} model; they were not used as feedback when calculating IK_{pc} UnSS kinematics. Mean (5 trials per participant), time normalized UnSS lower limb joint angles were calculated for the DK (criterion), IK_0 and IK_{PC} models.

Time-varying analyses of the UnSS lower limb kinematics were assessed over the stance phase of UnSS (vGRF > 10 N). Time-varying differences between the IK models and the established DK model were calculated using the open source one-dimensional statistical parametric mapping analysis package SPM1D{t} (α=0.05) (spm1d.org)[16] and root mean square errors (RMSE). Statistically significant differences were recorded as a percentage of stance and calculated using the average time in which the time-varying t-statistic breached the critical t-threshold. As a surrogate measure of effect size, RMSE were also calculated comparing the resulting kinematics of each modelling approach.

Results

The mean difference for all three DoF at the hip was 70±53% of stance (RMSE=14±11°, p<0.001 to 0.046) for the DK versus IK_0 comparison, and 46±40% of stance (RMSE=5±5°, p<0.001 to 0.038) for DK versus IK_{PC} comparison. For the hip flexion-extension (flex-ext) kinematics there were differences for 100% of stance (RMSE=18°, p<0.001) for DK versus IK_0 comparison (Fig. 1 a-b), and 41% of stance (RMSE=2.9°, p=0.016 to 0.030) for the DK versus IK_{PC} comparison (Fig. 1 g-h). Coincidentally, for the hip abd-add kinematics there were differences for the initial 9% of stance for both the DK versus IK_0 (RMSE= 2°, p=0.041), and DK versus IK_{PC} (RMSE = 2°, p=0.040) comparisons (Fig. 1 c-d, i-j). For the hip int-ext rotation kinematics there were statistical differences for 100% of stance (RMSE=23°, p<0.001) for the DK versus IK_0 comparison (Fig. 1 e-f), and the first 88% of stance (RMSE=11°, p<0.001) for the DK versus IK_{PC} comparison (Fig. 1 k-l).

<<Insert Fig. 1>>

The mean difference for all three DoF at the knee was 46±48% of stance (RMSE=8±7°, p<0.001 to 0.040) for the DK versus IK_0 comparison, and 56±31% of stance (RMSE=7±3°, p<0.001 to 0.046) for the DK versus IK_{PC} comparison. For the knee flex-ext there were differences for 7% of stance (RMSE=2°, p=0.038) for the DK versus IK_0 comparison (Fig. 2 a-b), and 22% of stance (RMSE=2°, p=0.006) for DK versus IK_{PC} comparison (Fig. 2 g-h). For knee abd-add there were differences for 100% of stance (RMSE=15°, p<0.001) for the DK versus IK_0 comparison (Fig. 2 c-d), and the last 83% of stance (RMSE=9°, p<0.001) for the DK versus IK_{PC} comparison (Fig. 2 i-j). For knee int-ext rotation there were differences for the initial 32% of stance (RMSE=8°, p=0.002) for the DK versus IK_0 comparison (Fig. 2 e-f), and the initial 64% of stance (RMSE = 10°, p=0.004 to 0.021) for the DK versus IK_{PC} comparison (Fig. 2 k-l).

<<Insert Fig. 2>>

Time-varying analysis of the ankle flex-ext kinematics showed differences for 100% of stance (RMSE=11°, p<0.001) for the DK versus IK_0 comparison (Fig. 3 a-b) and, 3% of stance (RMSE = 2°, p=0.044) for DK versus IK_{PC} (Fig. 3 c-d).

<<Insert Fig. 3>>

Discussion

This study aimed to investigate the measurement agreement of two different IK models; one prepared without prescribed participant-specific joint co-ordinates during the marker registration phase of model preparation (IK_0) and the other prepared with prescribed co-ordinates (IK_{PC}), against an established and inter- intra-tester repeatable DK model (criterion)[9,11,14,15,17]. Across all lower limb degrees of freedom analysed, it was found that prescribing joint co-ordinates during the marker registration phase of model preparation (IK_{PC}) resulted in kinematic outputs that more closely agreed with the kinematic estimates of the established DK model. This finding was substantiated by both time-varying statistical analyses and comparison to existing *in-vivo* research[18] (as well as discrete statistical analysis; Supplementary materials 2, Tables S1 & S2).

The prescription of participant-specific joint co-ordinates most influenced hip joint flex-ext kinematic estimates, with RMSE relative to the DK model in the magnitude of 18° for the IK_0 modelling approach and 3° for the IK_{PC} modelling approach. As mentioned previously, prescribing participant-specific joint co-ordinates during marker registration, the joint DoF within the model are aligned to the participant being tested[19]. This modelling step therefore reduces the potential for kinematic offsets from being introduced during model preparation, which would have significant downstream influences on IK derived UnSS lower limb joint kinematic estimates[19]. With such a large allowable flex-ext range of motion at the hip joint, these participant-specific postural offsets are likely attributed to the large RMSE observed when an IK_0 modelling approach was used. Interestingly, hip abd-add kinematics were in good agreement between the DK model and for both the IK_0 and IK_{PC} model estimates. Conversely, the int-ext rotation kinematics at the hip were in poor agreement with the DK model for both the IK_0 and IK_{PC} modelling approaches. In both instances, the DK model estimated significantly greater (RMSE=23°, p<0.001 and RMSE=11°, p<0.001 respectively) hip internal rotation kinematics across stance. This may, in part be explained by STA, which likely influenced the dynamic modelling of the pelvis' anatomical coordinate system differently when modelled with a DK versus IK modelling approach[7]. It is possible that the large upper body and pelvic movements typically observed during UnSS[15,17] influenced the IK derived int-ext rotation kinematics of the pelvis, and to a lesser degree, the abd-add DoF. The reason is because global optimisation computational procedures needs to fit all the inter-linked rigid segments of the rigid model as a whole to noisy experimental kinematic data[7]. As the purpose of this investigation was to assess the influence of model preparation on IK derived kinematics, not the mitigation of STA on the trunk and pelvis, a more through explanation pertaining to why these hip kinematic differences were observed needs to explored and verified with future research.

The prescription of participant-specific joint co-ordinates during marker registration did not influence the knee flex-ext kinematics, with the IK_0 and IK_{PC} derived estimates agreeing with the DK estimates (RMSE<2°). Conversely, IK_0 and IK_{PC} model hip int-ext rotation and knee abd-add angles were in poor agreement with the DK model. The DK model estimated knee adduction kinematics in the range of 10°- 25°, while the IK_0 model estimated peak knee abduction kinematics in the range of 0° - 10° and the IK_{PC} model in the range of 0° - 5° abduction. Interestingly, these results show that that IK_0 and IK_{PC} knee abd-add estimates are more physiologically plausible measurements of the underlying skeletal motion when compared to the DK model's estimates[20]. In addition IK results also align with those of Robinson et al.[9] and supported by *in-vivo* biplanar videoradiography data of Miranda et al.[18] who reported that approximately ± 5° of peak knee abd-add is observed during a jump-cut change of direction task. In the present study, the IK_{PC} model calculated knee joint kinematics most similar to the *in-vivo* data of Miranda et al.[18]. These findings together suggest that the prescription of participant- specific joint co-ordinates during marker registration is an important consideration for obtaining biologically feasible frontal plane knee kinematics during UnSS.

The prescription of participant-specific joint co-ordinates greatly influenced ankle plantar-dorsiflexion kinematic estimates. There was an 11° RMSE when comparing the DK model to the IK_0 model and only 2° RMSE when comparing the DK model to the IK_{PC} model. For researchers interested in estimating IK derived ankle joint kinematics, the prescription of participant-specific joint co-ordinates during the marker registration phase of model preparation is recommended.

A limitation of the present study was the use of an established inter- intra tester repeatable DK model. As DK models are known to be influenced by STA, particularity during high velocity sporting tasks, it is not possible to ascertain if the joint angle differences observed were entirely from measurement uncertainties from IK models or in part from the measurement uncertainties from the DK model. As a true 'ground truth' measure is currently not available, future research is recommended to build upon these findings. Ideally, the use of bone pins could be employed to assess accuracy; however, this is unlikely to be ethically plausible among healthy sport active individuals performing dynamic movement tasks like running, jumping and change of direction tasks. A less invasive alternative may be high-speed fluoroscopy, although current limitations of this approach include sampling rate restrictions and small motion capture volumes.

We hope this research is a step forward towards establishing modelling standards within the OpenSim modelling framework as the potential clinical benefits are vast. For example, with standardised model preparation procedures established, researchers and clinicians globally will be able to share and compare their clinical data and research with more confidence. This also allows for multi-centre, international prospective or clinical trials to operate with greater ease.

Conclusion

The prescription of participant-specific joint co-ordinates during the marker registration phase of model preparation is an important methodological consideration for obtaining biologically reasonable lower limb UnSS kinematics. We hope this research serves as a foundation for the establishment of standardised model preparation recommendations when using an inverse kinematic modelling approach.

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Figure Captions

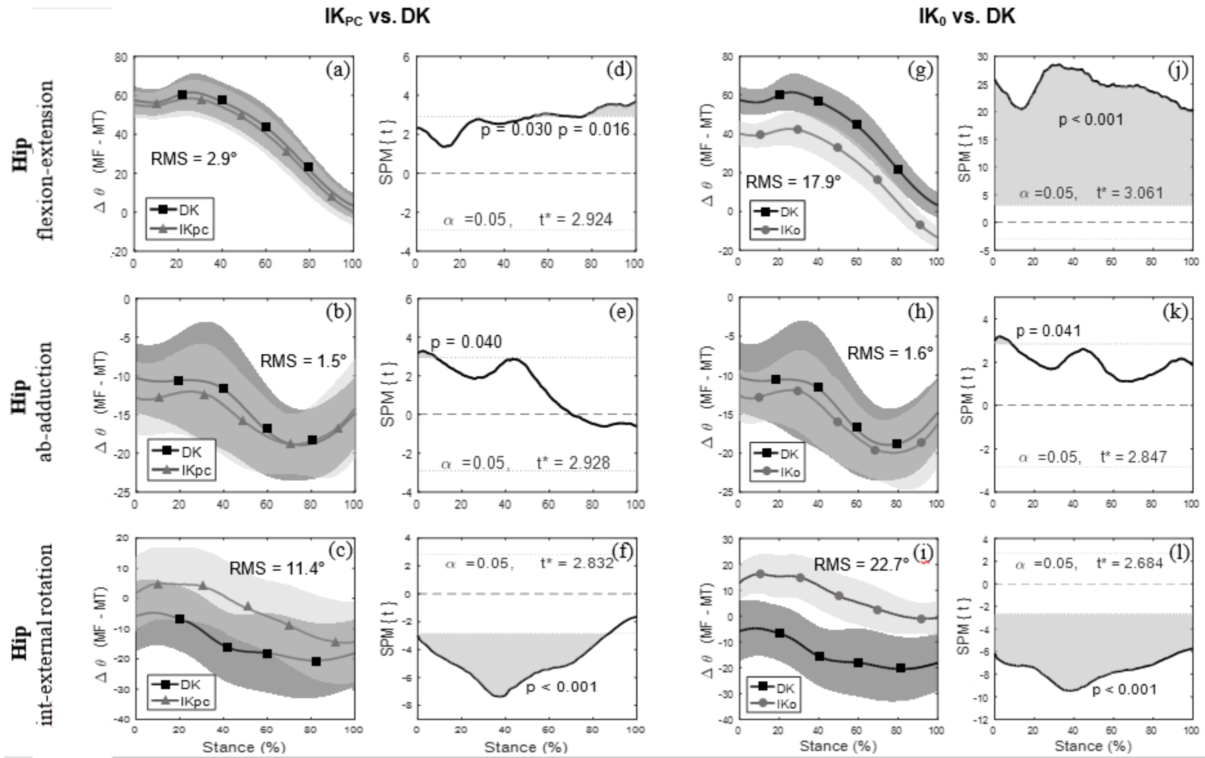

Fig. 1 Mean normalized hip angles (°) and RMSE values comparing DK versus IK_{PC} (a - c) and DK versus IK_0 (g - i). Graphs d - f, j - l are statistical parametric maps for the respective kinematic data. Shaded areas indicate significant differences between modelling approaches ($p < 0.05$). All curves are time normalized over the stance phase (%) for UnSS tasks.

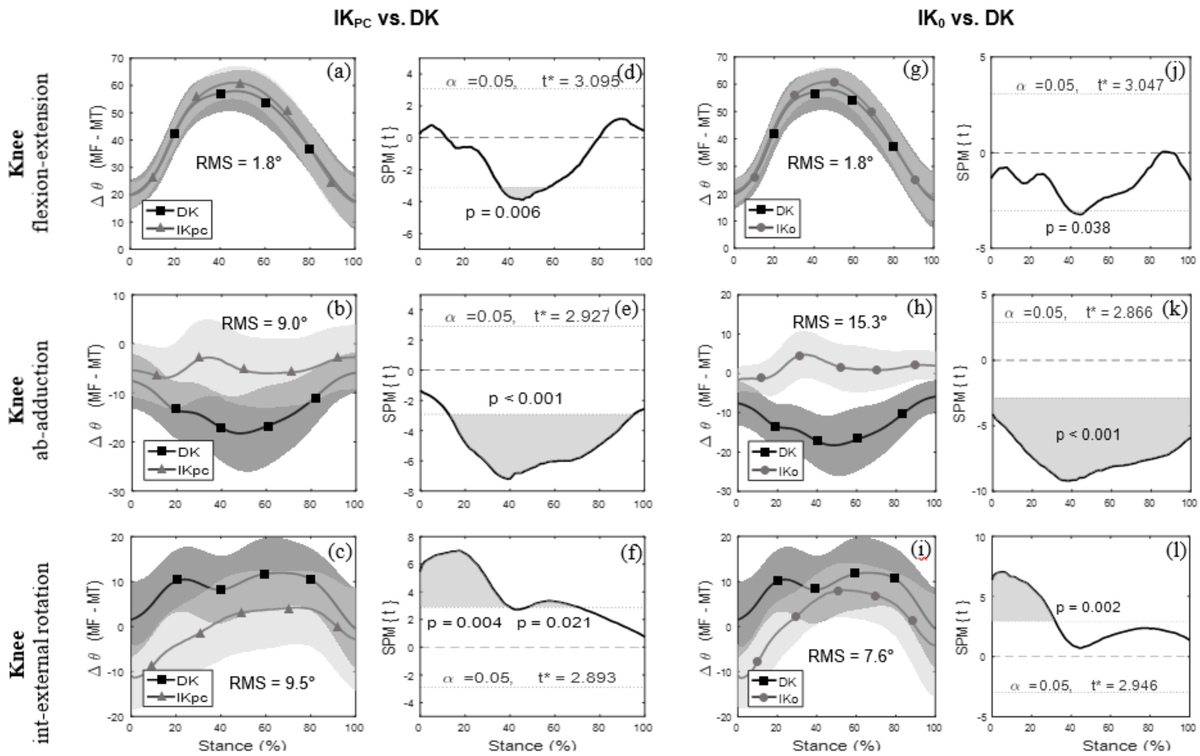

Fig. 2 Mean normalized knee angles (º) and RMSE values comparing DK versus IK_{PC} (a - c) and DK versus IK_0 (g - i). Graphs d - f, j - l are statistical parametric maps for the respective kinematic data. Shaded areas indicate significant differences between modelling approaches ($p < 0.05$). All curves are time normalized over the stance phase (%) for UnSS tasks.

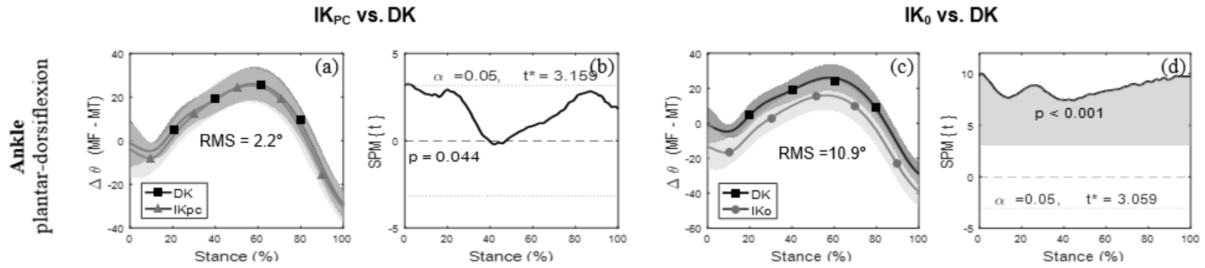

Fig. 3 Mean normalized ankle angles (º) and the RMSE value comparing DK versus IK_{PC} (a) and DK versus IK_0 (c). Graphs b, d are statistical parametric maps for the respective kinematic data. Shaded areas indicate significant differences between modelling approaches ($p < 0.05$). All curves are time normalized over the stance phase (%) for UnSS tasks.