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Kainz, H., Graham, D., Edwards, J. et al. (6 more authors) (2017) Reliability of four models for clinical gait analysis. Gait & Posture, 54. pp. 325-331. ISSN 0966-6362

https://doi.org/10.1016/j.gaitpost.2017.04.001

Accepted Manuscript

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PII:	S0966-6362(17)30113-3
DOI:	http://dx.doi.org/doi:10.1016/j.gaitpost.2017.04.001
Reference:	GAIPOS 5374

To appear in: *Gait & Posture*

Received date:	10-11-2016
Revised date:	31-3-2017
Accepted date:	1-4-2017

Please cite this article as: Kainz Hans, Graham David, Edwards Julie, Walsh Henry PJ, Maine Sheanna, Boyd Roslyn N, Lloyd David G, Modenese Luca, Carty Christopher P.Reliability of four models for clinical gait analysis.*Gait and Posture* http://dx.doi.org/10.1016/j.gaitpost.2017.04.001

Title page

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Reliability of four models for clinical gait analysis

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Research highlights

- Joint kinematic and kinetic reliability was high for all models
- Inverse Kinematic models are as reliable as the conventional gait model
- Musculoskeletal models are suitable for clinical gait analysis

Abstract

Three-dimensional gait analysis (3DGA) has become a common clinical tool for treatment planning in children with cerebral palsy (CP). Many clinical gait laboratories use the conventional gait analysis model (e.g. Plug-in-Gait model), which uses Direct Kinematics (DK) for joint kinematic calculations, whereas, musculoskeletal models, mainly used for research, use Inverse Kinematics (IK). Musculoskeletal IK models have the advantage of enabling additional analyses which might improve the clinical decision-making in children with CP. Before any new model can be used in a clinical setting, its reliability has to be evaluated and compared to a commonly used clinical gait model (e.g. Plug-in-Gait model) which was the purpose of this study. Two testers performed 3DGA in eleven CP and seven typically developing participants on two occasions. Intra- and inter-tester standard deviations (SD) and standard error of measurement (SEM) were used to compare the reliability of two DK models (Plug-in-Gait and a six degrees-of-freedom model solved using Vicon software) and two IK models (two modifications of 'gait2392' solved using OpenSim). All models showed good reliability (mean SEM of 3.0º over all analysed models and joint angles). Variations in joint kinetics were less in typically developed than in CP participants. The modified 'gait2392' model which included all the joint rotations commonly reported in clinical 3DGA, showed reasonable reliable joint kinematic and kinetic estimates, and allows additional musculoskeletal analysis on surgically adjustable parameters, e.g. muscle-tendon lengths, and, therefore, is a suitable model for clinical gait analysis.

1. Introduction

Children with cerebral palsy (CP) have complex musculoskeletal pathologies which are commonly corrected using single-event multilevel orthopaedic surgeries [1]. Three-dimensional gait analysis (3DGA) is used to inform the clinical decision-making in children with CP. Many clinical gait laboratories implement 3DGA methods that estimate joint kinematics and kinetics, but generally do not provide direct objective musculoskeletal information. The surgeon is therefore required to exercise a high level of clinical reasoning to extrapolate the results from 3DGA to develop a surgical plan. In recent years, user friendly musculoskeletal modelling software (e.g. OpenSim [2] and AnyBody [3]) has emerged that additionally enables calculation of muscle-tendon length [4], muscle moment arm [5] and joint contact forces [6]. The adoption of musculoskeletal modelling software for clinical

3DGA may provide additional data to identify musculoskeletal causes of dysfunction, thereby better informing the treatment decision-making process.

Many clinical gait laboratories rely on the conventional gait analysis model [7, 8], which employs a computational method termed Direct Kinematics (DK) to calculate joint kinematics. A commonly used variant of the conventional gait model is the Plug-in-Gait (PiG) model, available with the Vicon/Nexus software package. Our confidence in using the conventional gait model is in part due to the demonstrated reliability of kinematic and kinetic data, which suggests that the magnitude of the errors obtained using this model are clinically reasonable [9]. In addition to the conventional gait model, several modified DK models have been developed (e.g.[10]) that implement different marker configurations, anatomical and technical reference frames, joint constraints and/or joint rotation conventions. Many of these other models have similar reliability to the PiG model, but because PiG is a commonly used version of the conventional gait analysis model we selected PiG as a reference model in this study.

In contrast to DK models, musculoskeletal modelling software [2, 3] solve for Pose Estimation using Inverse Kinematics (IK), also known as global-optimization which has been demonstrated to reduce soft tissue artefacts [11]. Barriers for the widespread implementation of musculoskeletal models in clinical 3DGA are comparing the differences between commonly used DK and IK models, and determining the reliability of IK models compared to the conventional gait model. Regarding the latter, only a small number of studies have assessed the reliability of IK models [12-14]. A reliability study in a single healthy participant found significantly lower inter-tester variations in joint angles in an IK compared to the conventional DK model [12]. The reliability of knee kinematics and kinetics during the stance phase of side-cutting manoeuvers have been tested in healthy adults and no difference was found in the reliability of knee angles but a small reduction in the variability of joint moments in the sagittal and transversal planes in the IK model [14]. Similar reliability was found for DK and IK models when analysing pelvis, spine and lower limb movements during running in healthy adults [13]. To date, no studies have assessed the reliability of IK models in computing kinematic and kinetic gait profiles of typically developing (TD) children or of children with CP.

The aims of this study were to (1) determine the reliability of 3DGA kinematic and kinetic data using two IK and two DK models, and (2) quantify the differences in joint angles and net joint moments between the selected IK and DK models, with both aims being referenced to the PiG model. It was hypothesised that inter- and intra-tester reliability of gait kinematics and kinetics in participants with CP and TD participants would not differ between our IK and DK models and that differences between IK and DK models would not be significant.

2. Methods

2.1 Participants

Eleven participants with CP (4 female, 7 male, age: 10.3±4.0years, height: 1.33±0.16m, weight: 29.0±10.6kg, GMFCS level 1-3) and seven TD participants (3 female, 4 male, age: 12.5±3.6years, height: 1.47±0.16m, weight: 40.9±15.0kg) were recruited and presented for two data collection sessions. On the first session two gait analysts placed markers for 3DGA (for inter-tester comparisons). Both examiners had >10 years of experience in marker placement and conducting gait analysis. On the second session, approximately one week after the first session, one gait analyst performed a repeat 3DGA (for intra-tester comparison). Ethics approval was obtained from the

Queensland Children's Health Services Human Research Ethics Committee (HREC/13/QRCH/197) and all participants provided informed consent.

2.2 Motion capturing

Each gait analyst placed a superset of retro-reflective surface markers on each participant (Fig. 1) and, therefore, the same trials could be used for the calculation of joint kinematics and kinetics with all analysed models (described below). Marker trajectories of one static and at least six walking trials at a self-selected speed were collected at 100Hz using an eight-camera, three-dimensional motion capture system (Vicon Motion Systems, Oxford, UK). Ground reaction forces were simultaneously acquired at 1000Hz using three force platforms (AMTI, Waterdown, MA, USA). Vicon Nexus 1.8.5 (Vicon Motion Systems, Oxford, UK) was used to label and filter marker trajectories and filter force plate data, with filters being a Butterworth 4th order zero-lag dual-pass, low pass filter with a cut-off frequency of 6Hz.

2.3 Analysed models

Different anatomical reference frames and different pose estimation methods were used for the creation of the following models and calculation of kinematic and kinetic waveforms.

2.3.1 DK models

1. The Vicon Plug-in-Gait (PiG-DK) model [7, 8], a variant of the conventional gait analysis model, used DK to calculate joint kinematics and outputs three rotations for the pelvis segment, hip and knee joint and two rotations for the ankle joint. The hip joint centre in the PiG-DK model was defined using the pelvic marker locations and the leg length measure [8]. In accordance with standard clinical practices at the Queensland Children's Motion Analysis Laboratory (Brisbane, Australia), one static and one walking trial were collected and processed to evaluate the knee ab-/adduction kinematic profile, instead of using a knee alignment device. If the knee ab-/adduction profile exceeded a range of motion of 10° and exhibited cross-talk with knee flexion/extension, the thigh wand marker was adjusted and a new static trial was collected. This final static trial was used for all analysed models. A value for tibial torsion was not entered. All analyses were done in Vicon Nexus 1.8.5.
2. The six degrees-of-freedom (DoF) DK (6-DoF-DK) model was created using BodyBuilder software (Vicon Motion Systems, Oxford, UK). Marker positions from the static trial were used to define joint centres. The modified Harrington regression equations, using only pelvic width as a regressor [15-17], were used to define the hip joint centre. The knee joint centre was defined as the midpoint between the markers on the medial

and lateral epicondyles. The ankle joint centre was defined as the midpoint between the medial and lateral malleolus markers including a proximal-distal offset correction of 2.7% of shank length [18]. Pelvis and thigh anatomical coordinate systems (ACS) were created following the ISB recommendations [19]. The proximal-distal axis of the shank ACS was defined from the ankle joint centre to the knee joint centre. The medio-lateral axis was defined perpendicular to the longitudinal axis in plane with the lateral malleoli marker and the anterior-posterior axis was mutually perpendicular to the other two axes. The anterior-posterior axis of the foot ACS was defined from the heel to the toe marker. The medial-lateral axis was perpendicular to the previous axis in plane with a virtual point defined by the height adjusted 5th metatarsal head marker (height was set equal to the toe marker during the static pose). The proximal-distal axis was mutually perpendicular to the other two axes. Each ACS was stored in a technical coordinate system based on markers of the same segment (detailed information about the technical coordinate systems can be found in [20], Table 2). In the dynamic trials the ACS were reconstructed and used to calculate joint kinematics as Cardan angles with the flexion/extension-ab/adduction-internal/external rotation order between adjusted segments without imposing any joint constraints (similar to [21]) or using segment optimization pose estimation methods [11]. Pelvic rotations were calculated as Cardan angles between the pelvis ACS and laboratory coordinate system using the rotation-obliquity-tilt sequence [22]. Joint kinetics were calculated via Inverse Dynamics using segment mass and inertia parameters from [23].

2.3.2 IK models

1. The 'gait2392' OpenSim (3-1-1-DoF-IK) model [24] is a commonly used IK model, which allows three rotational DoF at the hip joint. At the knee joint it only includes one DoF (flexion-extension) and prescribes sagittal plane translation. The subtalar joint was locked, allowing only one DoF (dorsi-/plantarflexion) at the ankle and the torso was not included in this model.
2. The second IK model (3-3-2-DoF-IK) had a ball-and-socket joint (three rotational DoF, no translation between segments) at the hip and knee allowing knee ab-adduction and internal-external rotations additionally to the knee flexion-extension rotation in the 3-1-1-DoF-IK model. The subtalar joint was enabled allowing two

separate DoF for the ankle joint complex. Furthermore, the rotation-obliquity-tilt sequence [22] was used to calculate pelvis rotations and the pelvis ACS was modified to be in plane with the ASIS and PSIS anatomical landmarks/surface markers, similar to the PiG-DK model.

Both IK models were scaled to each person using scale factors derived from surface marker positions and joint centres [25] (supplementary Table S1). For the marker placer task in OpenSim, only the anatomical landmark markers were weighted, enabling the model's cluster markers to be adjusted according to the experimental marker locations (Fig. 1). During the IK task the pelvis, foot and cluster marker were weighted heavily (supplementary Table S2**).** All scaling, kinematic and kinetic analyses were done in OpenSim 3.2 [2].

Six participants with CP and one healthy participant did not include the 5th metatarsal head markers and therefore ankle angles and joint kinetics were not included in the 6-DoF-DK and 3-3-2-DoF-IK model for these participants.

2.4 Data analysis

Similar to the process used in the Queensland Children's Gait Laboratory (Brisbane, Australia), for each participant, session and model, all individual kinematic and kinetic waveforms were visualized, obviously erroneous traces were removed and mean waveforms were calculated using the same trials in all models. An average of five kinematic and four kinetic trials were used to calculate mean waveforms. Standard deviations (SD) and standard error of measurement (SEM), calculated as the root mean square average of the within participant SD [26], from intra- and inter-tester kinematic and kinetic waveforms were calculated in MATLAB (R2013a, The Math Works, Natick, USA) and used to assess the reliability of each model. Two-way repeated measures ANOVAs (models × participant groups) with simple contrast using the SD metric was used to evaluate if there were differences in the reliability between the PiG-DK and all other models. In the case of significant main effects, post-hoc comparisons were performed using Bonferroni corrections. The significance level was set to $p<0.05$. IBM SPSS Statistics 22 (IBM Corporation, New York, USA) was used to perform the ANOVAs. Differences of kinematic and kinetic waveforms between the PiG-DK and all other models were analysed using a one-dimensional statistical parametric mapping package (SPM1D [27]). A general linear model with repeated measure was used to evaluate if there are overall differences in waveforms between models. Post hoc scalar field t-test with Bonferroni correction was used to compute a statistical parametric map for each parameter and comparison. Additionally, root-mean-square-differences (RMSD) were reported.

3. Results

One TD and two CP participants could not join the second data collection session and therefore inter-tester reliability could only be obtained from these participants. In three participants with CP only joint kinematics were analysed because not enough clean force plate strikes could be collected across all sessions for the computation of joint kinetics.

The reliability of joint kinematics in the sagittal and coronal planes was similar between models with SEM below 4º (Fig. 2, Table 1). Intra-tester SD in the 3-1-1-DoF-IK and 3-3-2-DoF-IK models and inter-tester SD in the 3-1-1-DoF-IK model for hip internal-external rotations were significantly smaller ($p<0.05$) than in the PiG-DK model. Knee flexion-extension inter-tester SD were significantly smaller ($p=0.043$) in the PiG-DK than in the 6-DoF-DK model. Knee ab-adduction, knee and hip internal-external rotation and ankle dorsi-plantar flexion intra-tester SD were significantly smaller ($p<0.05$) in TD than CP participants. Maximum IK marker tracking errors were 2.3±0.7cm and 1.9±0.6cm for the 3-1-1-DoF-IK and 3-3-2-DoF-IK models, respectively.

SEM for the joint kinetics were below 0.08 Nm/kg for all models (Fig. 3, supplementary Table S3). Inter-tester SD for hip flexion-extension moments were significantly smaller ($p=0.005$) in the 3-3-2-DoF-IK than in the PiG-DK model and intra-tester SD for hip ab-adduction moments were significantly smaller ($p=0.007$) in the 6-DoF-DK than in the PiG-DK model. Intra-tester SD were significantly smaller ($p<0.05$) in TD than CP participants for hip flexion-extension and ab-adduction moments and inter-tester SD were significantly smaller ($p<0.05$) in TD participants for hip and knee flexion-extension moments and hip ab-adduction moments.

Kinematic waveforms obtained with the 6-DoF-DK and 3-3-2-DoF-IK were very similar to the PiG-DK results (Fig. 4, Fig. 5 and supplementary SPM1D results). Only significant difference between the 3-3-2-DoF-IK and PiG-DK model was observed for ankle plantar-/dorsiflexion from approximately 10% to 35% of the gait cycle. The 3-1-1-DoF-IK showed the largest differences to the PiG-DK model with mean RMSD above 10º and significant differences over the whole gait cycle for pelvic anterior/posterior tilt and hip flexion-extension angles. Kinetic waveforms were similar between the PiG-DK and all other models with significant differences only for hip and knee moments at sporadic time points (supplementary SPM1D results). RMSD in joint moments between the PiG and all other analysed models were below 0.6 Nm/kg.

4. Discussion

This study evaluated the reliability of kinematic and kinetic 3DGA output from two DK and two IK models, and quantified the differences between the DK and IK models with reference to the conventional gait model. Overall, reliability was high for all models. Both IK models had SDs below 5º for all joint angles, and SDs were mostly below 5º for the PiG-DK and 6-DoF-DK models, except for transverse plane joint angles. In agreement with our first hypothesis, the reliability of kinematic and kinetic outputs, except for transverse plane hip angles, were not significantly different between the PiG-DK model and both IK models. Finally, the kinematic and kinetic outputs from the 3-3-2-DoF-IK model were similar to the PiG-DK model, indicating that this model is suitable for computing joint kinematics and kinetics for clinical gait analysis.

Our reliability results for joint kinematics using IK are in agreement with [12] who reported an overall mean inter-tester SD of 2.4±1.1º for lower limb joint angles. Whereas, our results are in partial disagreement with [14] who found a higher SD for knee flexion-extension angles (up to 5.7º) for their IK model and also with [13] who found that lower limb joint kinematics in the transverse plane were slightly less reliable using IK compared to DK. Nonetheless, both of these studies were conducted on healthy athletes performing high intensity movements including sidestepping and running. Furthermore, [13] additionally included thoracic and lumbar spine segments in the IK analysis. Thus, inclusion of simplified spine segments and different study population might explain the discrepancy between the findings of [13] and our results.

The 3-1-1-DoF-IK model reliability results indicated lower inter- and intra-test SDs for hip internal-external rotation angles compared to the 3-3-2-DoF-IK model, suggesting that joint constraints, i.e. fewer degrees-of-freedom, might increase the reliability of joint kinematics when using IK. Therefore, research that focus predominately on healthy participants might benefit in terms of reliability from implementing a model with fewer degrees-of-freedom at the knee. The commonly used model in clinical 3DGA is however the conventional gait model, which outputs three rotations for the knee joint and, therefore, the 3-1-1-DoF-IK model would not be suitable in many clinical settings.

Our reliability results for joint kinematics using DK are in agreement with previous literature [9], displaying low SDs in the sagittal and frontal planes and higher SDs in the transverse plane. The PiG-DK model had the highest single SD value with a mean of 7.2º for intra-tester knee internal-external rotation angles and also the highest overall SD values with a mean of 3.2º across all joint angles. Mean SDs for the 6-DoF-DK, 3-1-1-DoF-IK and 3-3-2-DoF-IK were 2.7º, 2.2º and 2.6º, respectively, for all analysed joint angles. The relatively poor reliability of the PiG-DK model in the transverse plane is likely explained by the thigh and shank wand markers, which are used in the PiG-DK model to define the knee and ankle flexion-extension axes and unlikely caused by the fact that PiG-DK uses DK. Accurate and reliable placement of these wand marker is challenging [28] and errors in the definition of the knee flexion-extension axis can significantly impact on knee internal-external rotations [29]. Unlike the PiG-DK model, our 6-DoF-DK model defined ACS independently of the wand markers and showed on average smaller variations in joint kinematics (SD 2.7±2.0º) and kinetics (SD 0.033±0.016Nm/kg) than the PiG-DK model (SD 3.2±2.8º and 0.044±0.021Nm/kg), which confirmed the findings from [30].

In agreement with our second hypothesis, kinematic and kinetic waveforms obtained with the 6-DoF-DK and 3-3-2-DoF-IK models were similar to the PiG-DK model waveforms. The small differences between the PiG-DK and 6-DoF-DK/3-3-2-DoF-IK models were likely caused by the additional foot marker in the 6-DoF-DK/3-3-2-DoF-IK models, which enabled better 3D tracking of the foot segment, and due to the knee and ankle axes definitions being independent of the wand markers. The large differences for pelvic anterior/posterior tilt and hip flexion-extension angles between the PiG-DK and 3-1-1-DoF-IK models were caused by the different definition of the pelvic ACS as shown in our previous study [20]. [10] compared five different DK models and concluded that model conventions and definitions are more crucial than the chosen marker set. Our results confirmed their conclusion and further highlighted that the computational method (DK versus IK) has a minor impact on the kinematic and kinetic results.

This study has some potential limitations. First, different IK, 6-DoF-DK or models based on segment pose estimation [11] could lead to slightly different reliability results. Second, reliability was assessed in lean, young healthy people and children with CP and, therefore, the results cannot be generalized to other populations, especially not to obese people. Third, given the additional degrees-of-freedom in the 3-3-2-DoF-IK any analyses of muscle or joint contact forces or muscle induced accelerations would require inclusion of ligamentous constraints and refinement of muscle-tendon pathways and conditional via points. Fourth, Residual Reduction Analysis (RRA), an OpenSim tool often used to ensure dynamic consistency prior to musculoskeletal simulations, was not employed in this study, as our models did not include a torso segment. Fifth, only five participants with CP had sufficient markers on the foot to compute ankle internal/external rotations and therefore this measure should be interpreted with caution.

5. Conclusion

The 3-3-2-DoF-IK model showed mean SDs below 5º for all joint angles and included all the joint rotations currently reported in a clinical setting and therefore this model would be reasonable for clinical 3DGA. Furthermore, the 3-3-2-DoF-IK model allows additional musculoskeletal analysis, e.g. muscle-tendon lengths, which might improve clinical-decision making in children with CP.

Acknowlegments

This research was funded in part by a Children's Health Queensland Innovation Grant (to CPC, grant number 10386-4), a Griffith University Research Fellowship (to CPC, grant number 44141), a Griffith University Publication Assistance Scholarship (to HK) and an NHMRC Research Fellowship (to RNB).

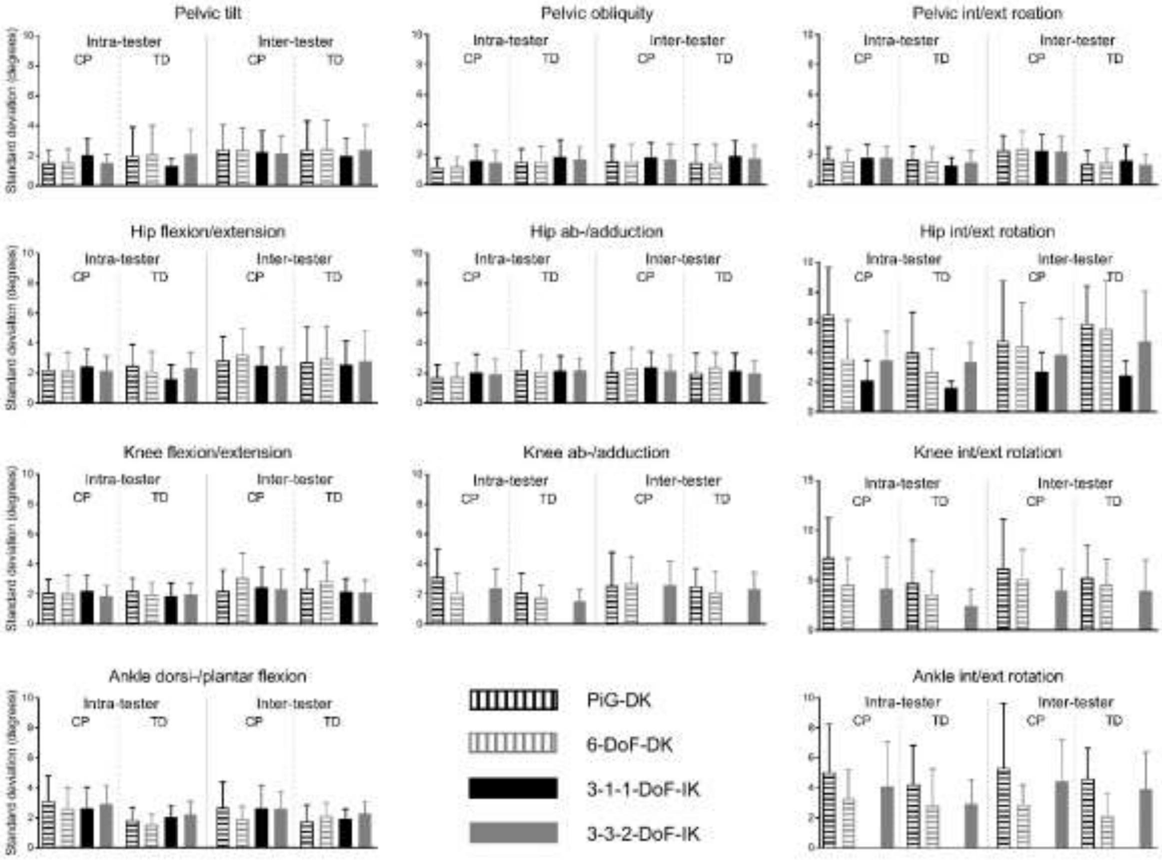
References

1. McGinley, J.L., et al., *Single-event multilevel surgery for children with cerebral palsy: a systematic review.* Dev Med Child Neurol, 2012. **54**(2): p. 117-28.
2. Delp, S.L., et al., *OpenSim: open-source software to create and analyze dynamic simulations of movement.* IEEE Trans Biomed Eng., 2007. **54**(11): p. 1940-50.
3. Damsgaard, M., et al., *Analysis of musculoskeletal systems in the AnyBody Modeling System.* Simulation Modelling Practice and Theory, 2006. **14**(8): p. 1100-1111.
4. Riley, P.O., et al., *Changes in hip joint muscle-tendon lengths with mode of locomotion.* Gait Posture, 2010. **31**(2): p. 279-83.
5. Arnold, A.S., et al., *Accuracy of muscle moment arms estimated from MRI-based musculoskeletal models of the lower extremity.* Comput Aided Surg, 2000. **5**(2): p. 108-19.
6. Saxby, D.J., et al., *Tibiofemoral contact forces during walking, running and sidestepping.* Gait Posture, 2016. **49**: p. 78-85.
7. Kadaba, M.P., H.K. Ramakrishnan, and M.E. Wootten, *Measurement of lower extremity kinematics during level walking.* J Orthop Res., 1990. **8**(3): p. 383-92.
8. Davis, R.B., et al., *A gait analysis data collection and reduction technique.* Human Movement Science, 1991. **10**(5): p. 575-587.
9. McGinley, J.L., et al., *The reliability of three-dimensional kinematic gait measurements: a systematic review.* Gait Posture, 2009. **29**(3): p. 360-9.
10. Ferrari, A., et al., *Quantitative comparison of five current protocols in gait analysis.* Gait Posture., 2008. **28**(2): p. 207-16. doi: 10.1016/j.gaitpost.2007.11.009. Epub 2008 Feb 15.
11. Lu, T.W. and J.J. O'Connor, *Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints.* J Biomech, 1999. **32**(2): p. 129-34.
12. Charlton, I.W., et al., *Repeatability of an optimised lower body model.* Gait Posture., 2004. **20**(2): p. 213-21.
13. Mason, D.L., et al., *Reproducibility of kinematic measures of the thoracic spine, lumbar spine and pelvis during fast running.* Gait Posture., 2016. **43:96-100.**(doi): p. 10.1016/j.gaitpost.2013.11.007. Epub 2014 Feb 2.
14. Sankey, S.P., et al., *How reliable are knee kinematics and kinetics during side-cutting manoeuvres?* Gait Posture., 2015. **41**(4): p. 905-11. doi: 10.1016/j.gaitpost.2015.03.014. Epub 2015 Mar 28.
15. Harrington, M.E., et al., *Prediction of the hip joint centre in adults, children, and patients with cerebral palsy based on magnetic resonance imaging.* J Biomech, 2007. **40**(3): p. 595-602. Epub 2006 Apr 3.
16. Sangeux, M., *On the implementation of predictive methods to locate the hip joint centres.* Gait Posture, 2015. **42**(3): p. 402-5.

17. Kainz, H., et al., *Estimation of the hip joint centre in human motion analysis: a systematic review.* Clin Biomech (Bristol, Avon), 2015. **30**(4): p. 319-29.
18. Bruening, D.A., A.N. Crewe, and F.L. Buczek, *A simple, anatomically based correction to the conventional ankle joint center.* Clin Biomech (Bristol, Avon), 2008. **23**(10): p. 1299-302.
19. Wu, G., et al., *ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion--part I: ankle, hip, and spine. International Society of Biomechanics.* J Biomech., 2002. **35**(4): p. 543-8.
20. Kainz, H., et al., *Joint kinematic calculation based on clinical direct kinematic versus inverse kinematic gait models* Journal of Biomechanics, 2016. **49**(9): p. 1658-69.
21. Leardini, A., et al., *A new anatomically based protocol for gait analysis in children.* Gait Posture, 2007. **26**(4): p. 560-71.
22. Baker, R., *Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms.* Gait Posture, 2001. **13**(1): p. 1-6.
23. de Leva, P., *Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters.* J Biomech, 1996. **29**(9): p. 1223-30.
24. Delp, S.L., et al., *An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures.* IEEE Trans Biomed Eng., 1990. **37**(8): p. 757-67.
25. Kainz, H., et al., *Accuracy and Reliability of Marker Based Approaches to Scale the Pelvis, Thigh and Shank Segments in Musculoskeletal Models.* Journal of Applied Biomechanics, 2017: p. 1-21.
26. Bland, J.M. and D.G. Altman, *Measurement error.* BMJ : British Medical Journal, 1996. **312**(7047): p. 1654-1654.
27. Pataky, T.C., M.A. Robinson, and J. Vanrenterghem, *Vector field statistical analysis of kinematic and force trajectories.* J Biomech., 2013. **46**(14): p. 2394-401. doi: 10.1016/j.jbiomech.2013.07.031. Epub 2013 Jul 31.
28. Baker, R., L. Finney, and J. Orr, *A new approach to determine the hip rotation profile from clinical gait analysis data.* Human Movement Science, 1999. **18**(5): p. 655-667.
29. Ramakrishnan, H.K. and M.P. Kadaba, *On the estimation of joint kinematics during gait.* J Biomech, 1991. **24**(10): p. 969-77.
30. Stief, F., et al., *Reliability and Accuracy in Three-Dimensional Gait Analysis: A Comparison of Two Lower Body Protocols.* Journal of Applied Biomechanics, 2013. **29**(1): p. 105-111.

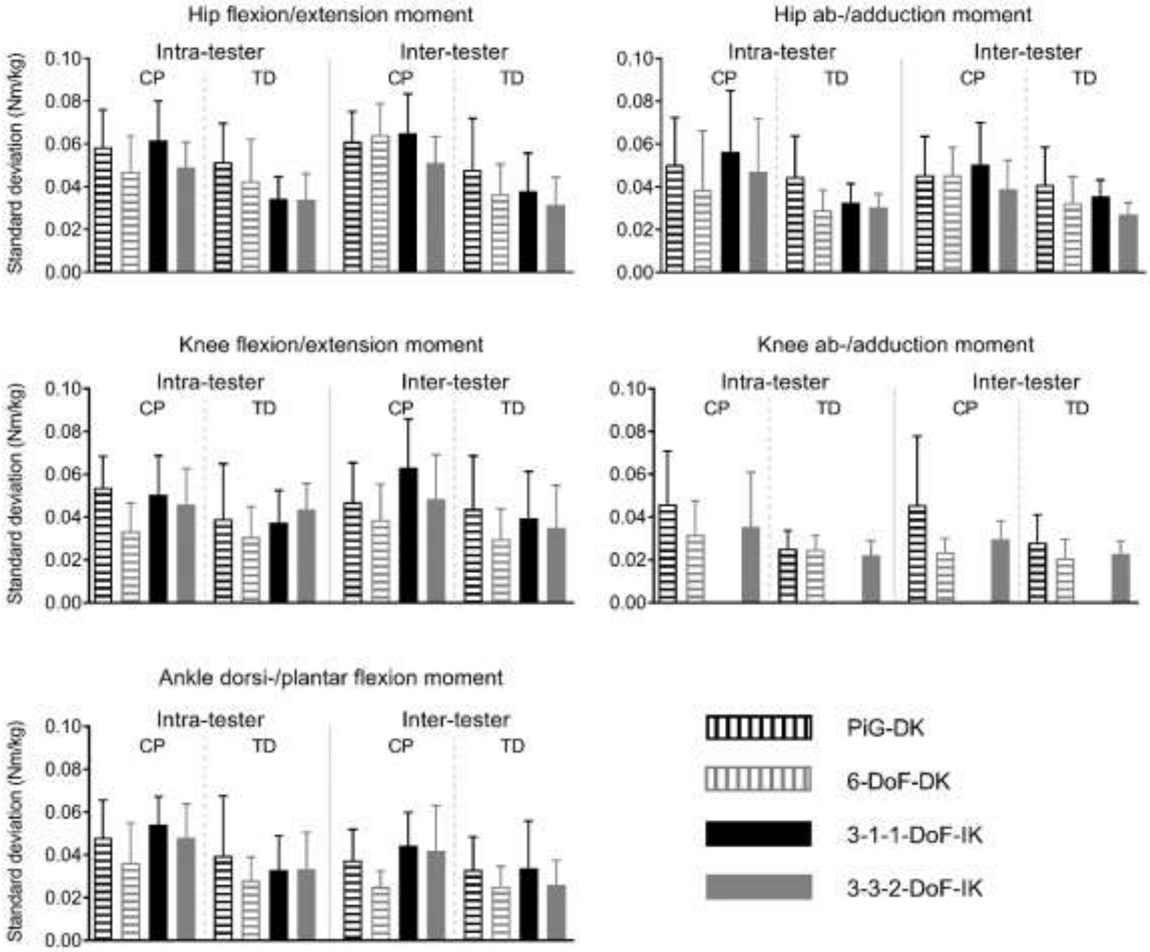
Figure Caption

Figr-1



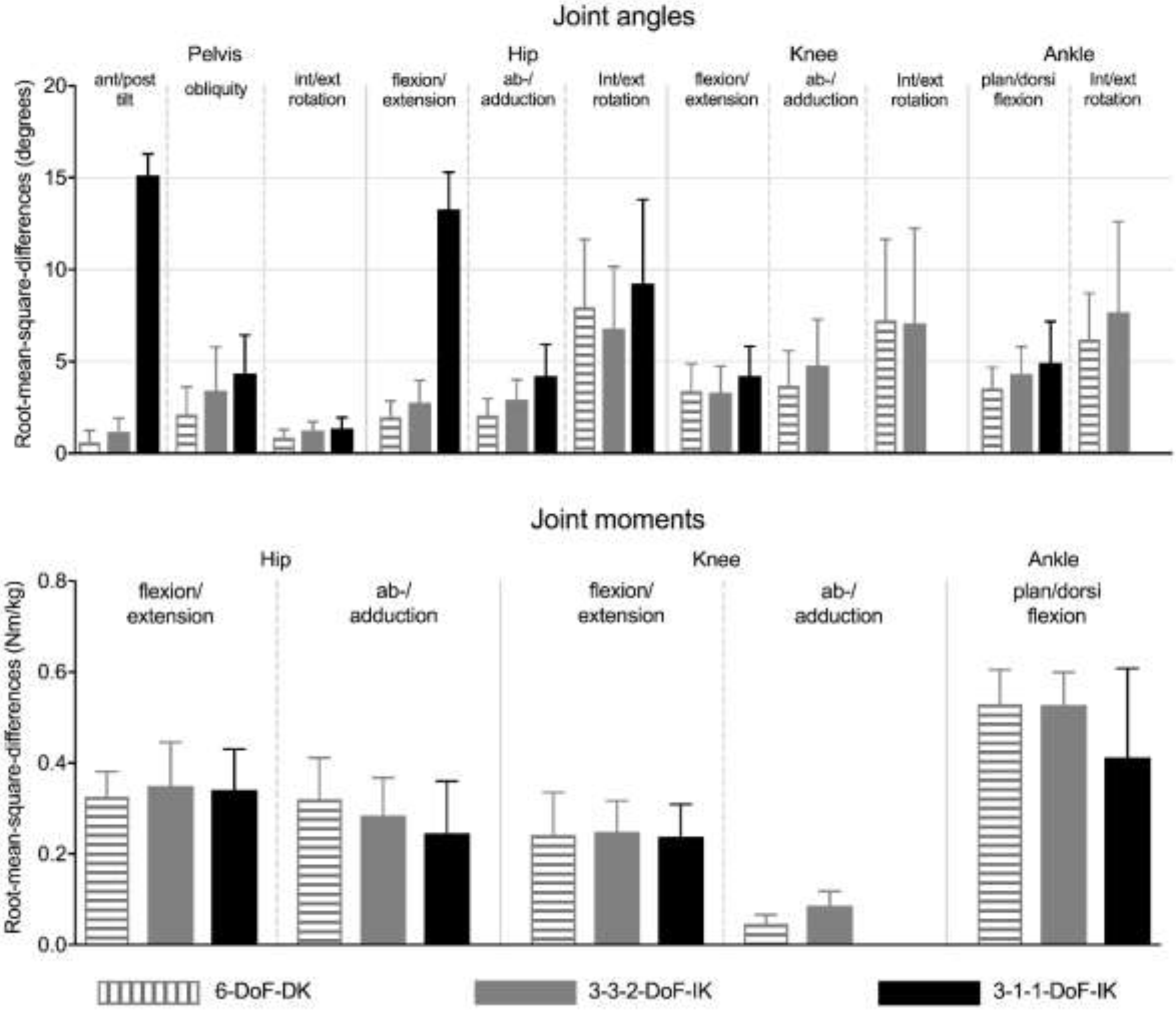
Pelvic tilt
Pelvic obliquity
Pelvic int/ext roation
Hip flexion/extension
Hip ab-/adduction
Hip int/ext rotation
Knee flexion/extension
Knee ab-/adduction
Knee int/ext rotation
Ankle dorsi-/plantar flexion
Ankle int/ext rotation
Intra-tester
Inter-tester
CP
TD
Standard deviation (degrees)
PiG-DK
6-DoF-DK
3-1-1-DoF-IK
3-3-2-DoF-IK

Figr-2

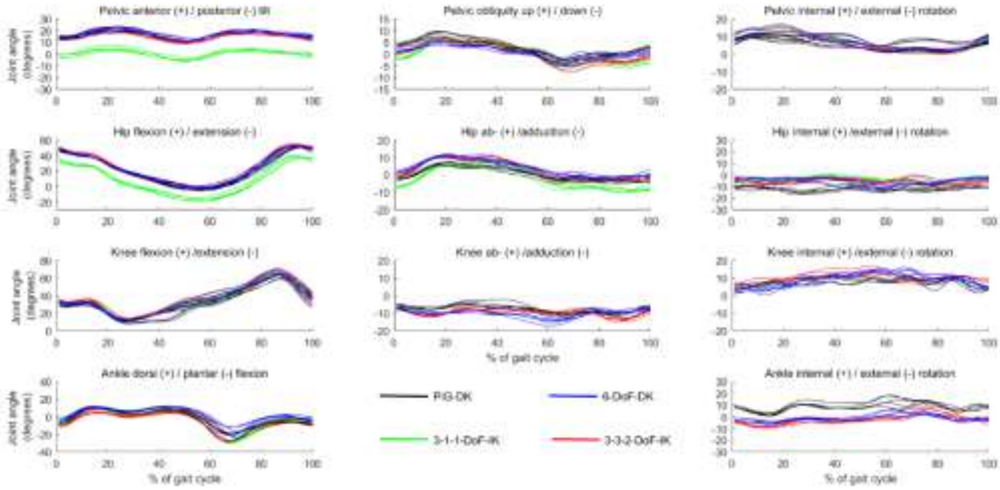


Hip flexion/extension moment
Hip ab-/adduction moment
Knee flexion/extension moment
Knee ab-/adduction moment
Ankle dorsi-/plantar flexion moment
Intra-tester
Inter-tester
CP
TD
Standard deviation (Nm/kg)
0.10
0.08
0.06
0.04
0.02
0.00
PiG-DK
6-DoF-DK
3-1-1-DoF-IK
3-3-2-DoF-IK

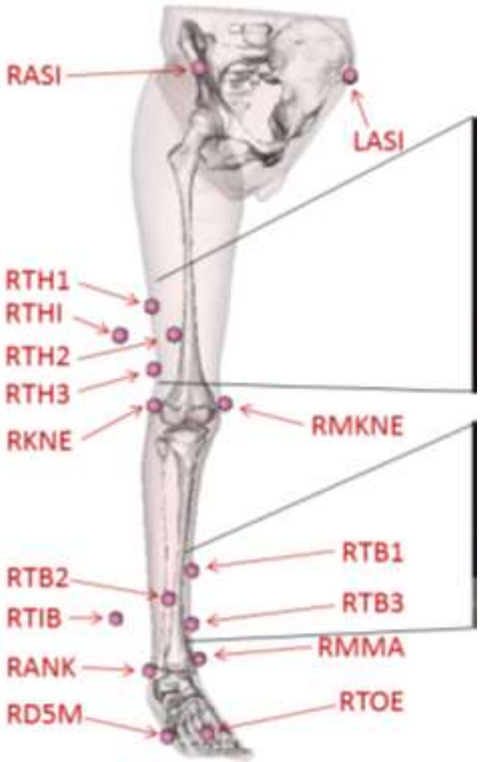
Figr-3


Joint angles
Pelvis
Hip
Knee
Ankle
ant/post tilt
obliquity
int/ext rotation
flexion/ extension
ab-/ adduction
Int/ext rotation
flexion/ extension
ab-/ adduction
Int/ext rotation
plan/dorsi flexion
Int/ext rotation
Root-mean-square-differences (degrees)
0
5
10
15
20
Joint moments
Hip
Knee
Ankle
flexion/ extension
ab-/ adduction
flexion/ extension
ab-/ adduction
plan/dorsi flexion
Root-mean-square-differences (Nm/kg)
0.0
0.2
0.4
0.6
0.8
6-DoF-DK
3-3-2-DoF-IK
3-1-1-DoF-IK

Figr-4


Pelvic anterior (+) / posterior (-) tilt
Pelvic obliquity up (+) / down (-)
Pelvic internal (+) / external (-) rotation
Hip flexion (+) / extension (-)
Hip ab- (+) /adduction (-)
Hip internal (+) /external (-) rotation
Knee flexion (+) /extension (-)
Knee ab- (+) /adduction (-)
Knee internal (+) /external (-) rotation
Ankle dorsi (+) / plantar (-) flexion
Ankle internal (+) / external (-) rotation
Joint angle (degrees)
% of gait cycle
PiG-DK
6-DoF-DK
3-1-1-DoF-IK
3-3-2-DoF-IK

Figr-5



Abbreviation	Placement/Full name	Model			
		PiG-DK	6-DoF-DK	3-1-1-DoF-IK	3-3-2-DoF-IK
RASI/LASI	anterior superior iliac spine	A, T	A, T	A, T	A, T
RPSI/LPSI	posterior superior iliac spine	A, T	A, T	A, T	A, T
RTHI/LTHI	thigh wand marker	T	-	-	-
RTH1/LTH1	thigh cluster marker 1	-	T	T	T
RTH2/LTH2	thigh cluster marker 2	-	T	T	T
RTH3/LTH3	thigh cluster marker 3	-	T	T	T
RKNE/LKNE	lateral knee	A, T	A	A, T	A, T
RMKNE/LMKNE	medial knee	A	A	A	A
RTIB/LTIB	shank wand marker	T	-	-	-
RTB1/LTB1	shank cluster marker 1	-	T	T	T
RTB2/LTB2	shank cluster marker 2	-	T	T	T
RTB3/LB3	shank cluster marker 3	-	T	T	T
RANK/LANK	lateral ankle	A, T	A	A, T	A, T
RMMA/LMMA	medial malleolus	A	A	A	A
RTOE/LTOE	Top of the second metatarsal head	A, T	A, T	A, T	A, T
RD5M/LD5M	Lateral at the head of the 5th metatarsal	-	A, T	A, T	A, T
RHEE/LHEE	Posterior aspect of the heel	A, T	A, T	A, T	A, T

Figr-8**Figure legends**

Fig. 1. Superset of surface markers placed on the participants and used in the PiG-DK, 6-DoF-DK, 3-1-1-DoF-IK and 3-3-2-DoF-IK models. The thigh cluster was placed lateral on the distal third of the thigh. The shank cluster marker was placed anterior on the distal third of each shank. The distance between markers on the long axis of the clusters was 10.5cm and the third marker of the clusters was perpendicular to the long axis 4.5cm from the midpoint. A = anatomical marker used to create the anatomical segment frames in the DK models or scale the generic IK models, T = tracking marker, - = marker was not used in this model. Only markers from the right leg are shown in the figure.

Fig. 2. Mean intra- and inter-tester standard deviation for kinematic waveforms obtained with the PiG-DK, 6-DoF-DK, 3-1-1-DoF-IK and 3-3-2-DoF-IK models. CP=participants with cerebral palsy. TD=typically developed participants. Error bars represent one standard deviation.

Fig. 3. Mean intra- and inter-tester standard deviations for kinetic waveforms obtained with the PiG-DK, 6-DoF-DK, 3-1-1-DoF-IK and 3-3-2-DoF-IK models. CP = participants with cerebral palsy. TD = typically developed participants. Error bars represent one standard deviation.

Fig. 4. Mean root-mean-square-differences (RMSD) for the comparison of joint angles and moments between the PiG-DK and 6-DoF-DK, 3-1-1-DoF-IK and 3-3-2-DoF-IK model. Error bars represent one standard deviation.

Fig. 5. Kinematic waveforms from 5 trials of one participant with cerebral palsy calculated with the PiG-DK, 6-DoF-DK, 3-1-1-DoF-IK, and 3-3-2-DoF-IK models. The same five walking trials were analysed in all models. The differences in pelvis and hip angles in the sagittal plane between the 3-1-1-DoF-IK and all other models were caused by the different anatomical segment frame definition at the pelvis segment as shown in our previous paper [20].

Table 1. Overall standard error of measurement (SEM) obtained by combining intra- and inter-tester standard deviations of kinematic waveforms calculated with the PiG-DK, 6-DoF-DK, 3-1-1-DoF-IK and 3-3-2-DoF-IK model. Within and between tester SEM for kinematic and kinetic waveforms can be found online in the supplementary Tables S3 and S4.

Joint angle	Participants with cerebral palsy				Typically developed participants			
	PiG-DK	6-DoF-DK	3-1-1-DoF-IK	3-3-2-DoF-IK	PiG-DK	6-DoF-DK	3-1-1-DoF-IK	3-3-2-DoF-IK
Pelvic tilt	2.4	2.4	2.5	2.1	2.8	2.9	1.9	2.7
Pelvic obliquity	1.6	1.7	2.0	1.8	1.8	1.8	2.1	1.9
Pelvic internal-external rotation	2.2	2.3	2.3	2.2	1.7	1.7	1.7	1.6
Hip flexion-extension	2.9	3.2	2.7	2.6	3.2	3.1	2.5	3.0
Hip ab-adduction	2.2	2.4	2.5	2.3	2.5	2.5	2.4	2.2
Hip internal-external rotation	6.6	4.8	2.8	4.2	5.7	5.1	2.2	4.8
Knee flexion-extension	2.4	3.0	2.6	2.4	2.5	2.7	2.2	2.2
Knee ab-adduction	3.5	2.9		2.9	2.6	2.3		2.2
Knee internal-external rotation	8.0	5.6		4.8	6.2	4.8		4.1
Ankle dorsi-plantar flexion	3.3	2.5	3.0	3.0	2.0	2.0	2.1	2.4
Ankle internal-external rotation	6.4	3.5		5.1	5.0	3.1		4.1