[2] N. Herssens, D. How, R. van de Berg, C. McCrum, Falls Among People With Bilateral Vestibulopathy: A Review of Causes, Incidence, Injuries, and Methods, JAMA Otolaryngol Head Neck Surg. 148 (2022) 187. https://doi.org/10.1001/jamaoto.2021.3673.
[3] A.R. Wu, C.S. Simpson, E.H.F. van Asseldonk, H. van der Kooij, A.J. Ijspeert, Mechanics of very slow human walking, Sci Rep. 9 (2019) 18079. https://doi.org/10.1038/s41598-019-54271-2.
[4] T. Brandt, M. Strupp, J. Benson, You are better off running than walking with acute vestibulopathy, The Lancet. 354 (1999) 746. https://doi.org/10.1016/S0140-6736(99)03179-7.

https://doi.org/10.1016/j.gaitpost.2023.07.087

Internal lower limb rotation increases patella cartilage pressure in individuals with patellofemoral instability

Bernhard Guggenberger [1,2,3], Brian Horsak [4], Andreas Habersack [1,5], Colin Smith [6], Martin Svehlik [1], Hans Kainz [2]

[1] *Department of Orthopaedics and Trauma, Medical University of Graz, Graz, Austria*
[2] *Neuromechanics Research Group, Department of Biomechanics- Kinesiology and Computer Science in Sport- Centre for Sport Science and University Sports, Vienna, Austria*
[3] *Institute of Physiotherapy, JOANNEUM University of Applied Sciences, Graz, Austria*
[4] *Center of Digital Health and Social Innovation, St. Pölten University of Applied Sciences, St. Pölten, Austria*
[5] *Institute of Human Movement Science, Sport and Health- University of Graz, Graz, Austria*
[6] *Department of Biomedical Engineering, Steadman Philippon Research Institute, Vail- CO, USA*

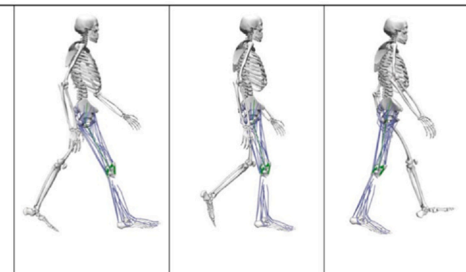
Introduction: Femoral anteversion (FA) and tibial torsion (TT) are known factors to contribute to patellofemoral instability [1]. Studies showed increased knee joint loading in individuals with idiopathic rotational deformities [2]. Investigating the effect of FA and TT on patellofemoral joint loading and lower leg muscle forces could help to better understand their impact on patellofemoral instability.

Research question: How do FA and TT influence the knee joint loading and muscle force in individuals with patellofemoral instability?

Methods: Musculoskeletal simulations were performed based on the retrospective data set of 11 individuals diagnosed with patellofemoral instability (13 affected knees). We measured FA and TT from magnetic resonance images. The mean FA and TT were $27 \pm 11^\circ$ and $31 \pm 8^\circ$, respectively. To account for the interaction of FA and TT, we calculated Lower Limb Rotation (LLR = TT-FA) [3]. The mean LLR value was $4 \pm 12^\circ$. For each individual an OpenSim model [4] was scaled for height, weight and maximum isometric muscle force [5]. We performed simulations for each participant based on a generic-scaled model and a second model with personalized FA and TT [6]. Kinematics, kinetics, muscle forces and patella cartilage pressure were estimated using the COMAK routine [7]. For muscle forces and patella cartilage loading, we calculated differences between both models for each participant and correlated them to LLR, FA and TT using Pearson correlation ($\alpha = 0.05$).

Results: LLR correlated to patella cartilage loading during different phases of stance (Fig. 1). No significant correlations were found between patella cartilage loading, FA and TT in our preliminary results. A more internally rotated LLR correlated with higher rectus femoris ($r = -0.79$, $p = 0.001$) and gluteus minimus force ($r = -0.77$, $p = 0.002$) as well as lower gluteus maximus ($r = 0.90$, $p < 0.001$) and gluteus medius force ($r = 0.71$, $p = 0.007$).

Fig. 1 - Correlation between LLR, FA, TA and patella cartilage pressure. Bold values are statistically significant ($p < 0.05$).



Gait cycle	Loading Response		Mid Stance		Terminal Stance and Pre Swing	
	Pearson r	**p-Value**	**Pearson r**	**p-Value**	**Pearson r**	**p-Value**
Lower Limb Rotation						
Maximum Pressure	-0.06	0.83	-0.53	0.06	**-0.64**	**0.02**
Average Pressure	-0.22	0.48	**-0.6**	**0.03**	-0.46	0.11
Contact Area	**-0.57**	**0.04**	**-0.75**	**0.003**	0.05	0.86
Femoral Anteversion						
Maximum Pressure	0.07	0.82	0.4	0.18	0.48	0.10
Average Pressure	0.29	0.34	0.19	0.53	0.19	0.54
Contact Area	0.32	0.29	0.51	0.08	0.12	0.686
Tibial Torsion						
Maximum Pressure	0.01	0.99	-0.20	0.50	-0.24	0.43
Average Pressure	0.08	0.79	-0.56	0.05	-0.38	0.20
Contact Area	-0.36	0.23	-0.36	0.23	0.23	0.44

Discussion: Our preliminary results showed that FA and TT solely did not correlate to patella cartilage loading. In contrast LLR rotated either internally or externally, changed patella cartilage loading. An explanation could be that higher LLR and thus greater differences between FA and TT increase the rotational limb malalignment and therefore loading at the patellofemoral joint. Considering the fact that patella dislocations commonly occur to the lateral side, internal rotation of the LLR might be more relevant, as it leads to a lateralization of the patella [9]. Therefore, individuals with internally rotated LLR might increase rectus femoris force, resulting in higher patella cartilage loading to stabilize the patellofemoral joint. Decreased forces of gluteal muscles might be explained by change of abductor lever arm caused by differences in position of the greater trochanter. To conclude, beside an individual interpretation of FA and TT, additionally LLR should be considered, when examining lower limb alignment in individuals with patellofemoral instability.

References

[1] W.R. Post, D.C. Fithian, Patellofemoral Instability: A Consensus Statement From the AOSSM/PFF Patellofemoral Instability Workshop, Orthopaedic Journal of Sports Medicine. 6 (2018) 232596711775035. https://doi.org/10.1177/2325967117750352.
[2] E. Passmore, H.K. Graham, M.G. Pandy, M. Sangeux, Hip- and patellofemoral-joint loading during gait are increased in children with idiopathic torsional deformities, Gait & Posture. 63 (2018) 228–235. https://doi.org/10.1016/j.gaitpost.2018.05.003.
[3] H.-A. Waidelich, W. Strecker, E. Schneider, Computertomographische Torsionswinkel- und Längenmessung an der unteren

Extremität, Fortschr Röntgenstr. 157 (1992) 245–251. https://doi.org/10.1055/s-2008-1033007.

[4] R.L. Lenhart, J. Kaiser, C.R. Smith, D.G. Thelen, Prediction and Validation of Load-Dependent Behavior of the Tibiofemoral and Patellofemoral Joints During Movement, Annals of Biomedical Engineering. 43 (2015) 2675–2685. https://doi.org/10.1007/s10439-015-1326-3.

[5] M.M. van der Krogt, W.W. de Graaf, C.T. Farley, C.T. Moritz, L.J. Richard Casius, M.F. Bobbert, Robust passive dynamics of the musculoskeletal system compensate for unexpected surface changes during human hopping, Journal of Applied Physiology. 107 (2009) 801–808. https://doi.org/10.1152/japplphysiol.91189.2008.

[6] K. Veerkamp, H. Kainz, B.A. Killen, H. Jónasdóttir, M.M. van der Krogt, Torsion Tool: An automated tool for personalising femoral and tibial geometries in OpenSim musculoskeletal models, Journal of Biomechanics. 125 (2021) 110589. https://doi.org/10.1016/j.jbiomech.2021.110589.

[7] C. Smith, R. Lenhart, J. Kaiser, M. Vignos, D. Thelen, Influence of Ligament Properties on Tibiofemoral Mechanics in Walking, Journal of Knee Surgery. 29 (2015) 099–106. https://doi.org/10.1055/s-0035-1558858.

[8] [8] T.C. Pataky, One-dimensional statistical parametric mapping in Python, Computer Methods in Biomechanics and Biomedical Engineering. 15 (2012) 295–301. https://doi.org/10.1080/10255842.2010.527837.

[9] A. Keshmiri, G. Maderbacher, C. Baier, F. Zeman, J. Grifka, H.R. Springorum, Significant influence of rotational limb alignment parameters on patellar kinematics: an in vitro study, Knee Surg Sports Traumatol Arthrosc. 24 (2016) 2407–2414. https://doi.org/10.1007/s00167-014-3434-2.

https://doi.org/10.1016/j.gaitpost.2023.07.088

Selective personalization of muscle-tendon properties for predictive simulations of walking in children with cerebral palsy

Dhruv Gupta [1], Bram Van Den Bosch [1], Ilse Jonkers [1], Anja Van Campenhout [2], Kaat Desloovere [3], Friedl De Groote [1]

[1] *KU Leuven, Department of Movement Sciences, Leuven, Belgium*
[2] *KU Leuven / UZ Leuven, Development and Regeneration, Leuven, Belgium*
[3] *KU Leuven / UZ Leuven, Rehabilitation Sciences, Leuven, Belgium*

Introduction: Changes in muscle-tendon (MT) properties contribute to gait deficits in children with cerebral palsy (CP) [1]. Predictive simulations of gait based on musculoskeletal (MSK) models can be used to study the effects of altered MT properties on gait mechanics and might inform treatment selection [1]. For accurate simulations, the MSK model should accurately capture the altered MT properties. These can be estimated by optimizing the fit between the simulated and experimental joint torques (inverse dynamics) [1]. To obtain reliable estimates, it is important that the joint torques are sensitive to the MT parameters. However, this is not sufficient, as MT parameters of synergistic muscles might have a similar effect on the joint torque. These factors limit the number of MT parameters that could be independently estimated from the available experimental data.

Research question: We aimed to study how reducing the set of estimated MT parameters based on their influence on joint torques (method A) and the similarities in their effect on joint torques (method B) influenced gait kinematics predicted using the resulting personalized MSK models.

Methods: For one child with CP (M, 12.8 yo, GMFCS II, pre-surgical assessment), we created a model (31 degrees of freedom, 82 Hill-type muscles) that accounted for alterations in musculoskeletal geometry based on MRI imaging [2]. We scaled MT parameters linearly before further personalization. We used data from walking (slow and fast), sit to stand, squat, counter movement jump, and instrumented passive spasticity assessment [3] to estimate MT parameters. First, we estimated optimal fiber and tendon slack length of all muscles [4] as independent variables by minimizing the squared difference between simulated and experimental joint torques. Second, we estimated the 75% MT parameters to which the simulated joint torques were most sensitive, while using the default values for the other MT parameters (method A). Third, we used techniques from system identification to determine linear combinations of MT parameters along with their effect on the cost function [5]. This method takes into account that different parameters might have a similar effect on the estimated joint torques. We estimated only 75% of the parameter combinations that had the highest effect on the cost function (method B). We then ran predictive simulations of gait using the MSK model with the linearly scaled MT parameters and the MT parameters determined from the mentioned methods.

Results: We found that estimating a reduced set of MT parameters (methods A and B) improved the agreement between the predicted and experimental gait pattern, especially for the knee (Fig. 1).

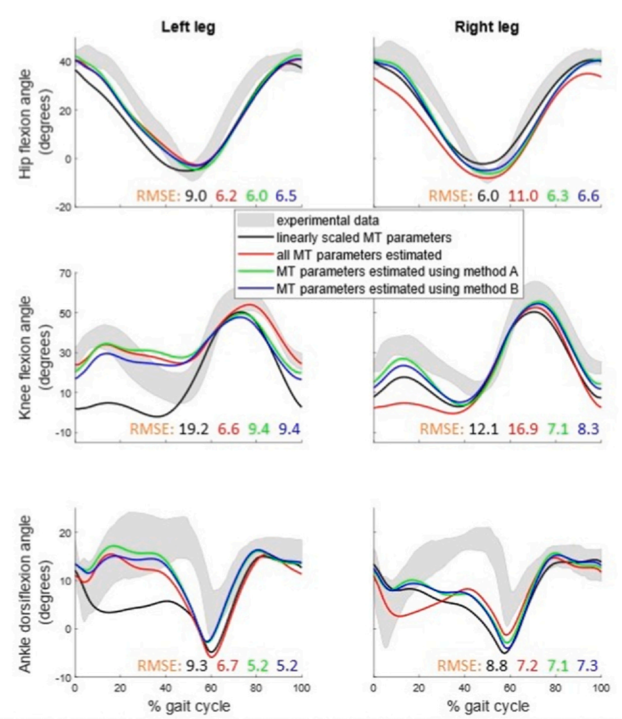


Figure 1: Sagittal plane kinematics from experimental data (mean ± 2 standard deviations) [grey], from predictive simulations of a musculoskeletal model using linearly scaled muscle-tendon parameters [black], all muscle-tendon parameters estimated and considered independent [red], only muscle-tendon parameters estimated to which the experimental joint torques were sensitive (method A) [green], muscle-tendon parameters estimated while taking into account that different parameters might have a similar effect on the estimated joint torques (method B) [blue]. The root mean squared error between the predicted kinematics and the mean experimental kinematics (RMSE) is also displayed. The crouch gait pattern in both knees is predicted only when using methods A or B.

Discussion: This study shows the importance of carefully determining which MT parameters can be estimated from the available