CENTER OF PRESSURE AND JOINT TORQUE ESTIMATION FOR SINGLE LEG SLACKLINE BALANCING USING MODEL-BASED OPTIMIZATION

Kevin Stein[1] and Katja Mombaur[1]

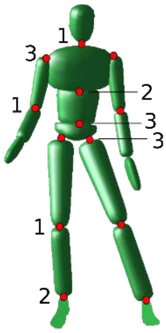
Optimization, Robotics and Biomechanics (ORB), Institute of Computer Engineering (ZITI), Heidelberg University, Heidelberg, Germany[1]

Being similar to tightrope walking, slacklining has become very popular among athletes and physiotherapists to practice and improve balancing capabilities. For flat ground static balance the center of pressure is often used to quantify how stable a subject is. In this work we present a method to reconstruct the center of pressure and the joint torques from pure motion capture data for motions that don’t allow for force plate measurements. We demonstrate the application to a subject balancing on a slackline. We create a subject-specific 3D-model and perform a least-squares fit to the recorded motion by formulation and solution of an optimal control problem. From the resulting forces we construct the center of pressure dynamics and quantify how stable the subject is on a slackline. The joint torques allow for further insight into the balancing strategies applied.

KEYWORDS: stability, multi-body modeling, center of pressure tracking

INTRODUCTION: Maintaining stability and balance is of crucial importance in human locomotion. Even though it is not fully understood how stability and balance are to be quantified, the center of pressure (CoP) position and velocity have been used in several studies to compare the balance capabilities of different athletes, for example by Thompson, Badache, Cale, Behera and Zhang (2017). Balancing and walking on a slackline is widely applied for balance training in various sports such as climbing or surfing but also in rehabilitation. The athlete hereby balances on a up to 5 cm wide nylon webbing that is tensioned between two fixed end points. Donath, Roth, Rueegge, Groppa, Zahner and Faude (2013) have performed CoP measurements standing on a force plate before and after slackline training to find transition effects. They found no significant improvement in static and dynamic stance. Yet no direct measurements of the CoP position while balancing on a slackline are reported in the literature. This is mainly due to that it would require force plates or pressure sensitive soles to acquire this data. Felis, Mombaur & Berthoz (2015) used model-based optimization to gain more insight into human walking identifying the whole dynamics of the model including the ground reaction forces using only kinematic data. Following their approach, but with a modified contact model, we present a method to reconstruct the CoP position and velocity during single leg balancing on a slackline from motion capture data. Additionally we can estimate the joint torques of the subject which allows for further interpretation on the balancing strategies applied.

METHODS: The method can be divided in three parts: modeling, motion capture and optimization. To describe balancing on a slackline with enough detail to estimate the CoP we need a subject-specific 3D dynamic model. For this purpose we rely on the anthropometric measurements by de Leva (1996) to estimate the segment geometry, masses and inertia. The model consists of 16 segments (head, upper and lower trunk, pelvis, upper and forearms, upper and lower legs, feet and hands). They are connected by joints which we model with the following degrees of freedom (DoF). The pelvis is linked to the world frame by three translational and three rotational DoF. To match the mechanical capabilities of a human, the hips, shoulders, and lower spine have spherical joints with 3 DoF. The ankles and the upper spine have 2 DoF and elbow, neck and knees have 1 DoF. The hands are fixed with respect to the forearms to further reduce the model complexity as they are not significant to the motion. In total the model has 32 DoF. Out of these 32 DoF 26 are actuated by torques that describe the effect of the accumulated muscle force per joint.



Segment deLeva	Joint	DoF wrt. parent
Pelvis	Floating Base	6 DoF
Lower Trunk	Lower Spine	RX, RY, RZ
Upper Trunk	Upper Spine	RY, RZ
Head	Neck	RY
Upper Arm	Shoulder	RX, RY, RZ
Fore Arm	Ellbow	RY
Hand	Wrist	Fixed
Upper Leg	Hip	RX, RY, RZ
Lower Leg	Knee	RY
Foot	Ankle	RX, RY

Figure 1: Left: Overview of the degrees of freedom (DoF) used for the deLeva Model. The Floating Base is not actuated and has three rotational and three translational DoF. The actuated rotational joints of the model are indicated with R and X, Y and Z describe the according coordinate system axis. Right: A typical slackline situation in the motion capture lab. The slackline is 5cm wide and 3m long and installed with the Gibbon Slackrack 300.

The model and an overview of the DoF is shown in Figure 1. We define the X-axis along the slackline, such that both form a right-handed system with the vertical Z-axis. Since this model is still highly complex we use the Rigid Body Dynamics Library (RBDL, Felis, 2017) to compute the kinematic and dynamic properties needed for the method.

The contact with the slackline is modeled by four contact points at the foot of the model such that a rectangular contact area of 5 x 20 cm is described. Two points are located at the heel and two at the hallux of the models foot. At each point a force can act on the model. Each force is described as a three dimensional vector that consists two tangential components and a normal component.

Motion was recorded with a camera-based motion capture system (Qualisys, Göteborg, Sweden) and the corresponding software (Qualisys Track Manager). The subject was prepared with 46 infrared-reflective markers following the Gate-IOR marker-set by Leardini, Biagi, Merlo, Belvedere and Benedetti (2011). The Slackline (5 cm wide and 3 m long), was attached at the two end points, but could otherwise move in space, using a Gibbon Slackrack 300. For generation of reference joint angles from the motion capture data we used the tool Puppeteer (Felis et al. 2015).

The last step of the method was the formulation and solution of an optimal control problem (OCP). We search for forces at the previously defined contact points and joint torques that make the model most closely track the fitted joint angles. Therefore the objective function minimized the squared deviation between the reference and the models joint angles. To avoid the use of excessive torques we added a small regulation term on the joint torques. Since all four contact forces are acting on the same body, this lead to an overdetermined problem formulation, which we solved by adding another small regularization term that also minimized the squared forces. As constraints to the OCP we used the kinematic limitation of the joint angles and the torque ranges that are reported in the literature for example by King, Wilson and Yeadon (2006). Further, we imposed a non-slip condition by introducing a friction coefficient that constraints the tangential component of the force with respect to the normal component. An OCP was solved by the direct multiple shooting method as implemented in the optimal control code MUSCOD-II (Bock & Plitt, 1984; Leineweber, Bauer, Bock & Schlöder, 2003). As stated in the introduction, the advantage of this approach compared to a classical inverse dynamics approach is that it does not require force plate measurements but allows us to reconstruct full dynamic model properties from purely kinematic measurements.

From the four calculated contact forces we were able to compute the CoP displacement from the center of the foot. We used the normal component of each force as described in Winter (2009). The displacement was computed individually for the mediolateral (ML, along the coronal plane) and anterior-posterior component (AP, along the sagittal plane).

RESULTS AND DISCUSSION: As a proof of concept, we applied the method to two recordings of the same subject on a slackline for 20 s on each leg. The average fitting error of the markers to the model was 1.36 cm. The OCP found a dynamically feasible motion that very closely tracked the reference motion with a deviation of 0.36 cm for the floating base and 0.2° for the models joint angles. From the resulting contact forces, we computed the CoP position and velocity over time for the AP and ML direction. The results were evaluated following the work of Thompson et al. From the position we derived the distance between the minimum and maximum AP and ML position (MAXD) and the root mean square (RMS). From the velocity we computed the mean (MV) and also the root mean square (RMSV). For all values a smaller number is favorable as we assumed that a perfectly balance subject is able to maintain the CoP always at the same spot. The results are shown in Table 1. The RMS values of the positions are within the range reported in the literature for two and single leg balancing for example by Mansfield and Innes (2015) and suggest that the subject was maintaining balance well. The MAXD, MV, and RMSV are higher than what is reported for standing on force plates, which is reasonable since the task is more difficult and the subject constantly has to adjust the pose to maintain balance. Similar values are observed for ML and AP direction.

Table 1: Evaluation of the reconstructed CoP properties

	Right		**Left**	
	AP	ML	AP	ML
RMS [cm]	0.27	0.28	0.29	0.29
MAXD [cm]	2.36	2.12	2.42	2.42
MV [cm/s]	1.64	1.74	2.45	1.86
RMSV [cm/s]	3.39	3.99	4.83	3.29

Further, the OCP results give us information on the joint torques that were applied by the subject to perform the motion. We find continuous activation and torques acting on the supporting leg and the two spine joints to maintain an upright posture. Out of the remaining limb joints the hip and and shoulders show the highest torques and activation. These are plotted in Figure 2 for balancing on the right foot.

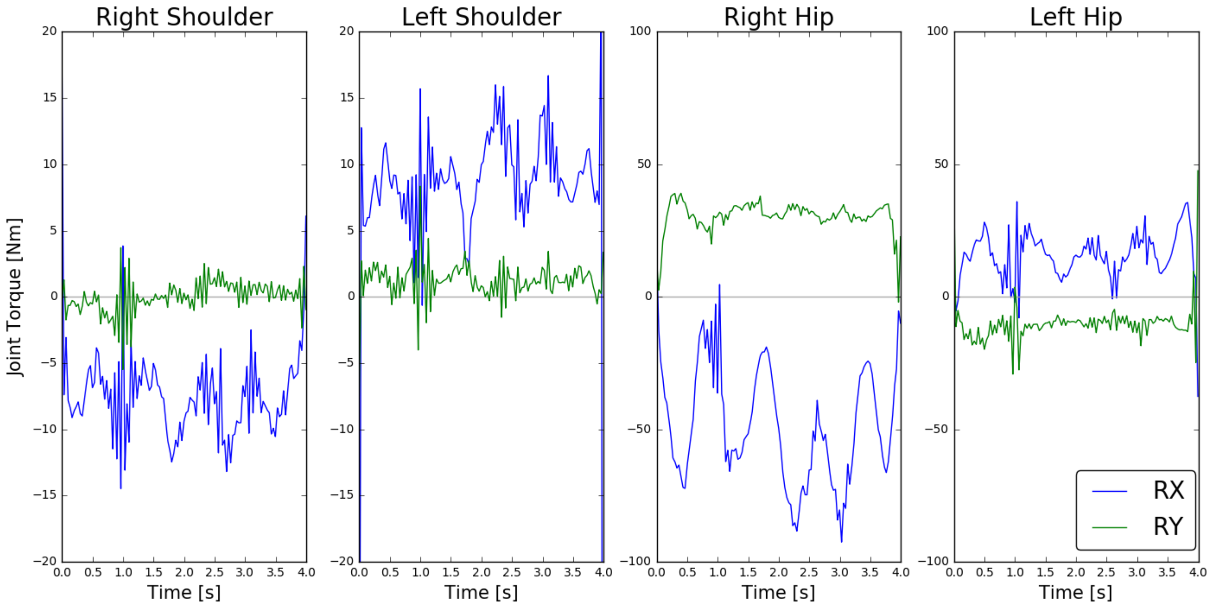


Figure 2: The resulting joint torques of the model for hip and shoulders joints. Higher actuation is seen in the coronal plane (RX) than for the sagittal plane (RY). Balance in ML direction is maintained by using the arms and legs, while the AP is balanced with the supporting leg's hip joint.

When separating the individual DoF according to their axis we find that for the coronal plane (RX-joint, plotted in blue) the actuation is generally higher than for the the sagittal plane (RY-joints, plotted in green). This supports the fact, that balance in ML direction (perpendicular to the slackline) is harder to maintain than in AP direction (along the slackline). Especially the shoulder joints are mainly creating torques around the X-axis to control balancing in the ML direction. For balancing in AP direction the main control takes place in the supporting legs hip joint. The whole body is leaning slightly forward or backward with the free leg balancing against the upper body.

CONCLUSION: The method allowed us to reconstruct the CoP position and dynamics from pure motion capture data and for situations in which no force plates can be used. Additionally, we get the model joint torques that were applied by the subject to perform the motion. We demonstrated the application to balancing on a slackline and computed the dynamics of the CoP to quantify how stable a subject is. We found similar values to what is reported in the literature. Analyzing the resulting joint torques we found different balance strategies for coronal and sagittal plane. In the future we are going to apply this method to compare the stability of different subjects on a slackline and also to monitor their learning process.

REFERENCES

Bock, H.G. & Plitt, K.-J. (1984). A multiple shooting algorithm for direct solution of optimal control problems. *Proceedings of the 9 th IFAC World Conference, Budapest*, 242-247.

Donath, L., Roth, R., Rueegge, A., Groppa, M., Zahner, L., & Faude, O. (2013), Effects of Slackline Training on Balance, Jump Performance & Muscle Activity in Young Children, *International Journal Of Sports Medicine*

Felis, M.L., Mombaur, K. & Berthoz, A. (2015). An optimal control approach to reconstruct human gait dynamics from kinematic data. *IEEE/RAS International Conference on Humanoid Robots*, 1044-1051.

Felis, M.L. (2017). RBDL: an efficient rigid-body dynamics library using recursive algorithms. *Autonomous Robots*, 41, 495-511.

King, M.A., Wilson, C., Yeadon, M.R. (2006). Evaluation of a torque-driven model of jumping for height,” *Journal of Applied Biomechanics*, vol. 22, no. 4, pp. 264–274, 2006.

Leardini, A., Biagi, F., Merlo, A., Belvedere, C., Benedetti, M.G. (2011). Multi-segment trunk kinematics during locomotion and elementary exercises, *Clinical Biomechanics*, Volume 26, Issue 6.

Leineweber, D.B., Bauer, I., Bock, H.G. & Schlöder, J.P. (2003). An Efficient Multiple Shooting Based Reduced SQP *Strategy for Large-Scale Dynamic Process Optimization (Parts I and II). Computers & Chemical Engineering*, 27, 157-174.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*, 29, 1223-1230.

Mansfield, A., & Inness, E. (2015). Force plate assessment of quiet standing balance control: perspectives on clinical application within stroke rehabilitation. *Rehabilitation Process and Outcome*.

Thompson, L.A., Badache, M., Cale, S., Behera, L., Zhang, N. (2017). Balance Performance as Observed by Center-of-Pressure Parameter Characteristics in Male Soccer Athletes and Non-Athletes. *SPORTS 2017*, 5, 86.

Winter, D.A. (2009) Biomechanics and Motor Control of Human Movement, Fourth Edition, ISBN: 978-0-470-39818-0.

Acknowledgement: We want to thank the Simulation and Optimization research group of the IWR at Heidelberg University for giving us the possibility to work with MUSCOD-II. We want to thank Carl Zeiss for funding the motion capture lab.