MODELING AND OPTIMAL CONTROL OF ABLE-BODIED AND UNILATERAL AMPUTEE RUNNING

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The remarkable performances of amputee athletes in sprint competitions aroused media and scientific interest and led to the question whether running-specific prostheses can be an advantage with respect to able-bodied running. The aim of this study was to bring together motion capture data and Scientific Computing methods to analyze the running motions of an able-bodied and a unilateral transtibial amputee athlete. For each of them a rigid multibody system model was created. By application of optimal control techniques, the dynamics of reference running movements from motion capture data was reconstructed for both models. The able-bodied and the transtibial amputee sprinters rely on dissimilar actuation strategies to perform similar running motions.

KEY WORDS: athletics, multibody system modeling, optimization, torque, prosthesis.

INTRODUCTION: Due to the greatly improving performances of amputee athletes in the last years, the research on running-specific prostheses and amputee running increased. Brüggemann, Arampatzis, Emrich and Potthast (2008) compared the biomechanics of a bilateral transtibial amputee and able-bodied athletes for sprinting at maximum speed. Their claim was that the amputee relies on a completely different motion pattern than able-bodied sprinters due to differences in the ground reaction forces and the ankle joint moments. This was confirmed by Weyand, Bundle, McGowan, Grabowski, Brown, Kram and Herr (2009). The point-counterpoint article (Weyand & Bundle, 2010; McGowan, Grabowski, Brown, Kram & Herr, 2010) points out the difficulty of generalizing any findings as only few disabled athletes compete at world-level. Wank and Keppler (2015) highlight that it is unfeasible to exactly specify any net (dis)advantage because an amputee athlete cannot be compared to himself without the amputation and his sensory impairments cannot be quantified. Mombaur (2014) used optimal control techniques to study sprinting of able-bodied and bilateral transtibial amputee athletes providing a way of comparing an amputee athlete to an able-bodied one of comparable figure. Therefore the purpose of this study was to approach the nature of unilateral transtibial amputee running by means of Scientific Computing methods and use optimization methods to further analyze practical motion capture recordings.

METHODS: The results presented in this paper mainly rely on the use of efficient multibody system models, motion capture and state of the art optimal control techniques.
The first pillar of our methods are efficient multibody system modeling tools. To describe human running motions, we use two different models, one of a unilateral transtibial amputee sprinter and one of an able-bodied athlete of comparable figure. Both models consist of 14 segments (head, upper and lower arms, three torso segments, thighs, shanks, feet/prosthetic device) with 16 degrees of freedom (DOF). As the motion is restricted to the sagittal plane, three DOF are associated with the overall position and orientation and the remaining ones describe the rotations of the internal joints. The internal DOF are powered by joint torques which we assume to summarize the action of all related muscles. In the amputee case, the prosthetic device is modeled by rigid components with a rotational joint and coupled to the remaining part of the shank by a fixed joint. It does not comprise an actuator, but a linear spring-damper system. The spring and damping constants are free parameters which need to be determined by optimization. For the creation of a subject-specific model, the de Leva data (de Leva, 1996) were extrapolated to the heights of 1.80m (able-bodied subject) or

1.83m (amputee subject), the overall weights of 75.4kg (able-bodied subject) or 76.0kg (amputee subject) and the measured segment lengths (see figure 1).

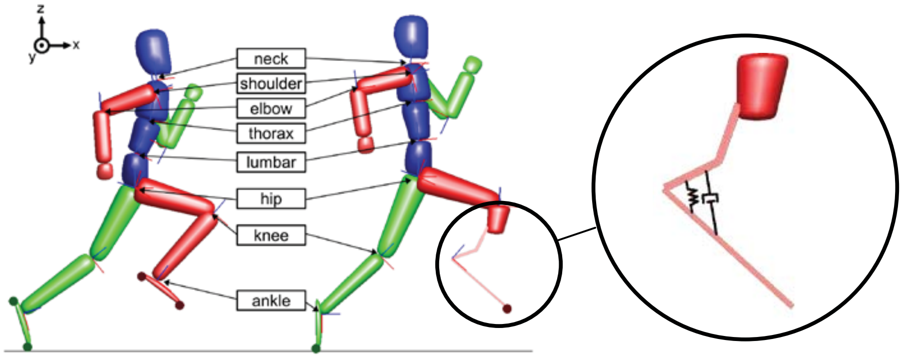

Figure 1: Models of an able-bodied and a unilateral transtibial amputee athlete

Running motions are characterized by a sequence of alternating flight and single-leg contact phases each of which is described by its own set of differential equations. The forefoot running is modeled by a point-like contact with the ball of the foot which we assume to be rigid and non-sliding. Thus the touch-down of the foot is completely inelastic, i.e. the velocity of the contact point is instantly set to zero, resulting in velocity discontinuities. As models of such complexity cannot be derived by hand, we use the tool RBDL (Felis, 2017).

The second part of our methods is the technique of capturing motions. The athletes performed runs at maximum speed on an indoor athletics track at the German Sport University Cologne. Their motions were recorded by a 3D camera system (VICON TM, Oxford, UK) comprising 16 infrared cameras operating at 250 Hz. For the recordings, retro-reflective markers were placed on anatomic landmarks and the prosthetic device using adhesive tape. For the generation of reference joint angles from the motion capture data we use the tool Puppeteer (Felis, Mombaur & Berthoz, 2015).

Thirdly, to conclude the elaboration of our methods for studying running motions, we introduce optimal control problems. To reconstruct the dynamics of the reference movements from motion capture data, we formulate and solve a multi-phase least squares optimal control problem. The objective function minimizes the deviations between the joint angles of the reference movement and the model. As constraints we use the multi-phase mechanical model as well as all kinematic and dynamic limitations of the model and the device such as joint angle and torque limits. Such problems can be solved by the direct multiple shooting method as implemented in the optimal control code MUSCOD-II (Bock & Plitt, 1984; Leineweber, Bauer, Bock & Schlöder, 2003). The advantage of this optimal control based approach compared to a classical inverse dynamics approach is that it does not require force plate measurements but allows to reconstruct full dynamic model properties from purely kinematic measurements. In addition, the fit to all markers can be taken into account in a balanced way, and there is no unfavorable error propagation along the kinematic chain.

RESULTS AND DISCUSSION: We successfully reconstructed the dynamics for both the able-bodied and the amputee running motions. The average values of the absolute errors are below 1cm for the translational and below 1.40° for the rotational DOF. Figure 2 shows animation sequences of the optimized solutions for both models. Figures 3a and 3b contain the histories of the joint angle and torque variables for both legs. Since the times for the completion of two full steps differ (t_{AB} = 0.444s , t_{AMP} = 0.448s), we normalized the swing and stance phase times to make them comparable. The phases (flight – right leg contact – flight – left leg contact) are separated by dashed lines. The contact phases are highlighted in grey.

The histories of the joint angles show similar shapes, they are however slightly shifted against each other. This is especially true for the curves of the right leg which is the affected leg in the amputee case. Both subjects have a comparable maximum hip extension, but the amputee reaches it within the second flight phase as opposed to the able-bodied's maximum hip extension at the end of the contact phase. It is remarkable that the knee angle flexion of

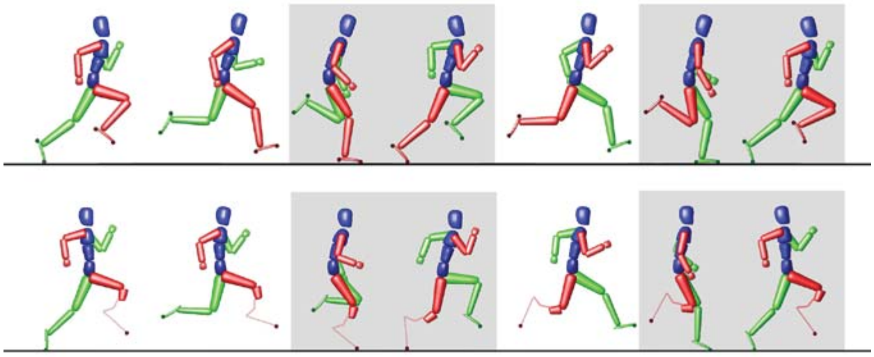

Figure 2: Animation sequences of optimized able-bodied and unilateral amputee running

the amputee is much smaller compared to the one in able-bodied running. The histories for the amputee's left leg differ only slightly from those of the able-bodied athlete. Hence, the joint angle trajectories suggest that the motions are not too different on the whole.

On the level of joint torques though, there are remarkable differences between the two of them. The joint torques in the left legs are of comparable size. However, the ones in the right hip and knee are clearly smaller in the amputee than in the able-bodied case, except for the right leg contact phase. Within this phase, the shape of the right leg joint torques is closely related to the corresponding shape of the horizontal and vertical ground reaction forces (GRF); we find similarities between the knee joint torque and the horizontal GRF as well as between the ankle joint torque and the vertical GRF for a biological leg. As there is no muscular torque in the prosthetic ankle, $\tau_{ankle} = 0$ for the amputee's right leg and the generation of vertical GRF is transferred partly to the hip and partly to the knee joints. Taken as a whole, it appears that some of the work actively done in able-bodied running is compensated for by the passive action of the spring-like prosthetic device, but within the respective contact phase the amputee has to do more work in the hip and knee joints.

In the case of a unilateral amputee athlete, one conceivable way towards a judgement is to investigate the asymmetry of left and right steps. As there is already a height difference between the amputee's left and right leg, we expected to find a considerable asymmetry in his running pattern. This was confirmed by the computational results, even though neither the able-bodied athlete runs as periodic nor the amputee athlete runs as asymmetric as anticipated. This becomes evident by considering the diagrams in figure 3c which show the differences in the joint angles of the legs between the left and the right step. For the calculation we subtracted the first step from the second one taking into account that the function of the limbs is interchanged, i.e. we distinguish stance and swing leg. The differences between the two steps are smaller in the able-bodied case as the absolute differences are maximally 23° as opposed to 57° in the amputee case. Even if we ignore the asymmetry in the ankle joints, the differences in the amputee's running motion are almost twice as big as the able-bodied ones. We thus deduce that the amputee running motion as a whole is less periodic than the able-bodied one. As we analyzed just two trials, the results might be specific to the respective athlete. Thereby we aim to extend our study to more subjects such that we are able to make more general statements.

CONCLUSION: The results suggest that able-bodied and amputee athletes use different strategies on the motor level to carry out similar running motions. Furthermore we were able to show that the unilateral amputee running motion is less periodic than the able-bodied one. We are still quite far from answering the question whether running-specific prostheses might provide the user an advantage over able-bodied athletes. However, we demonstrated that dynamic models and optimal control techniques can be a useful tool to reconstruct and investigate motions and provide information about non-measurable quantities such as torque histories. Future computations will consider more trials as well as other motions such as accelerated running and long-jump.

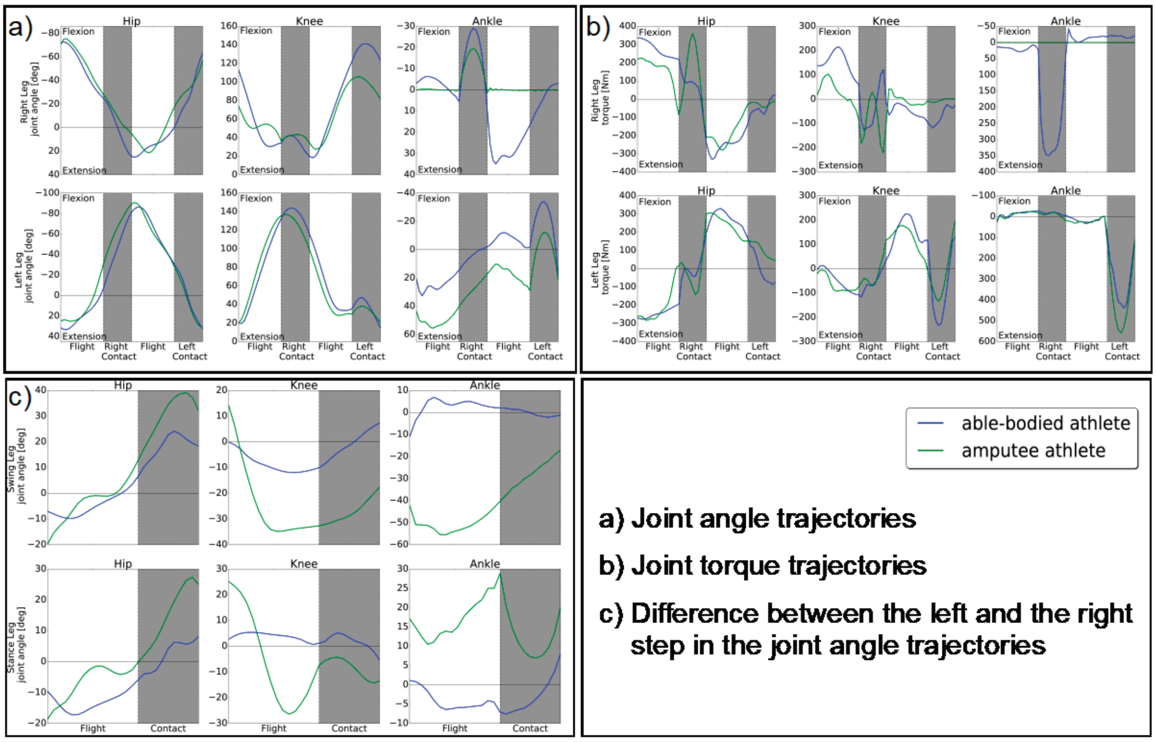

Figure 3: Optimized solutions of the leg joints for running with and without prosthesis

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