Prediction of muscle forces in residual limb during walking: comparison of transfemoral and Gritti-Stokes amputations

N. Fougeron[a,b]*, X. Bonnet[a], B. Panhelleux [a,c], J-L. Rose [b], P-Y. Rohan[a] and H. Pillet[a]

[a] *Institut de Biomécanique Humaine Georges Charpak, Arts et Métiers Paristech, 151 Boulevard de l'Hôpital, 75013 Paris, France;* [b] *Proteor, Recherche et développement, 5 boulevard Winston Churchill, 21000 Dijon, France;* [c]*Department of Surgery and Cancer, Imperial College London, 89 wood lane, W12 0BZ, London, United Kingdom*

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1. Introduction

Radcliffe's principle of lateral stabilization (Radcliffe 1955) was the theoretical basis of many socket designs. It describes the balance of the weight, generating a moment at the hip, of the amputated person thanks to the socket design. This principle was summarised in (Pritham 1990). Considering subischial sockets, the pelvic tilt induced by this moment is stabilised when the femur is abducted and pressures the lateral wall of the socket thanks to the gluteus medius (GM) action. Yet, this often leads to painful shear stresses in the medial tissues of the thigh. On the contrary, ischial containment sockets (ICS) rely on an ischial support to compensate the pelvis tilt which reduces the need for the GM activation. However, this support is often pointed out to be uncomfortable. In ICS, the frontal balance of the amputated person is a trade-off between the force applied on the support and the forces of the muscles. Evaluation of muscle forces is thus crucial to improve people's stability and comfort in their prosthesis. This could be achieved with musculoskeletal (MSK) modelling. However, few 3-dimensionnal models have been developed for transfemoral amputation in the literature. OpenSim models (Seth et al. 2018) were developed for that purpose (Bae et al. 2007; Mohamed 2018). Nevertheless, none of these models considered the vertical action applied by the ischial support. However, as emphasized by (Pritham 1990) it can be hypothesized that more than 83 % of the amputated person's weight is carried by the ischial support. Therefore, the moment at the hip induced by the ischial support cannot be neglected. This study presents a method to predict muscle forces in the residual limb from inverse dynamic and static optimisation from an MSK model. This method was applied on one transfemoral (TF) amputated person with different percentages of loading on the ischial support to evaluate its influence on muscle force estimation. The results were also compared to one Gritti-Stokes (GS) amputated subject for whom there is no ischial support. Muscles are also less impaired in GS amputations.

2. Methods

2.1 Data Acquisition

One right GS (BMI=27.4 cm.kg^{-2}) and one left TF (BMI=26 cm.kg^{-2}) amputated subjects were involved in this study. Both underwent amputation more than seven years prior to this study. The protocol combined motion analysis data and whole-body EOS radiography in the upright position (EOS®, EOS-Imaging, France) following the protocol described in (Al-Abiad et al. 2020). Mean values of kinematic and dynamic variables were computed for the optimization.

2.2 Musculoskeletal Modelling

The MSK model was implemented on Matlab (MATLAB R2014b The MathWorks, Inc., Natick, Massachusetts, United States). The OpenSim model Gait 2354_simbody was used to define generic muscle insertions in bones anatomical frames and generic maximal isometric forces per muscle. Insertions that were distal to the amputation level were placed at the extremity of the residual femur.

2.3 Personalisation of MSK models

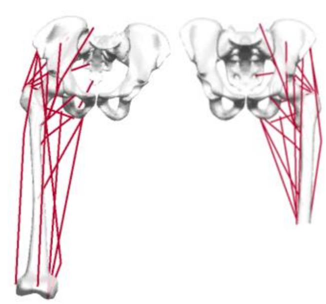

Figure 1: Subject-specific musculoskeletal models

EOS reconstructions were used to define personalized bone geometries (Mitton et al. 2006; Chaibi et al. 2012). Subject-specific origins and pathways of muscles were computed from kriging of generic muscle points using bone mesh nodes as control points (Figure 1).

2.4 Static optimisation

Static optimisation was based on the method proposed by (Son et al. 2012). The objective function is presented in Eq. 1 and the conditions to be satisfied in Eq. 2.

Eq. 1 $J(x) = \sum_{i=1}^{n}\left(\frac{F_i}{F_i^{max}}\right)^2$

Eq. 2 $$\begin{cases} \begin{pmatrix} r_{x1} & \cdots & r_{xn} \\ r_{y1} & \cdots & r_{yn} \\ r_{z1} & \cdots & r_{zn} \end{pmatrix} \times x = \begin{bmatrix} M_x \\ M_y \\ M_z \end{bmatrix} \\ 0 \leq x \leq F^{max} \end{cases}$$

Where n is the number of muscles, F_i the force of the i-th muscle, F^{max} the maximal isometric force, r the muscular lever arms and M the moment at the hip deduced from inverse dynamics. Lever arms were computed using effective origins and insertions of muscles as in (Zajac & Gordon 1989). The authors hypothesized no ischial support for the GS subject while this support was assumed to decrease the adduction/abduction moment by 83, 90 and 100 % for the TF subject. Prediction with no ischial support (0 %) was also computed for comparison purposes.

3. Results and discussion

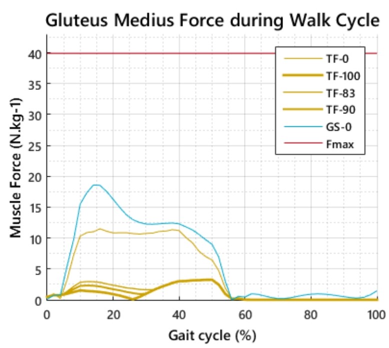

Figure 2: GM forces during a gait cycle.

Results are only detailed for the GM as it is the main muscle responsible for the lateral. Differences in the strategy of walking are noticeable comparing both types of amputation with 0 % ischial support since the GM is more recruited by the GS subject (Figure 2). The influence of the ischial support is also noticeable for the TF subject: the lower the adduction moment, the lower the GM action. As explained earlier: the tilting of the pelvis must be balanced by the femur abduction through the GM action. Moreover, as the support under the ischial tuberosity increases in ICS, the need for abduction is reduced. Pattern and force of activation are supported by electromyographic (EMG) studies (Wentink et al. 2013).

4. Conclusions

Evaluation of muscle forces is relevant to understand walking strategies of amputated subjects. Such results could be implemented in finite element modelling to study the interaction between the residual limb and the socket by adding muscles to the model to improve pressure prediction at the interface. Further studies are needed to accurately estimate the percentage of vertical loads supported by the ischial support and to evaluate the results with, for example, EMG data.

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***Corresponding author. Email: nolwenn.fougeron@ensam.eu**