To cite this version :

Nolwenn FOUGERON, Xavier BONNET, Brieuc PANHELLEUX, Jean-Loïc ROSE, Pierre-Yves ROHAN, Hélène PILLET, Helene PILLET - Effect of the ischial support on muscle force estimation during transfemoral walking - Prosthetics & Orthotics International - Vol. 49, n°1, p.p.51-59 - 2024

Effect of the ischial support on muscle force estimation during transfemoral walking

Fougeron, Nolwenn[1,2]; Bonnet, Xavier[1]; Panhelleux, Brieuc[3]; Rose, Jean-Loïc[2]; Rohan, Pierre-Yves[1]; Pillet, Hélène[1]

Abstract

Transmission of loads between the prosthetic socket and the residual limb is critical for the comfort and walking ability of people with transfemoral amputation. This transmission is mainly determined by the socket tightening, muscle forces and socket ischial support. However, numerical investigations of the amputated gait, using modelling approaches such as MusculoSkeletal (MSK) modelling, ignore the weight-bearing role of the ischial support. This simplification may lead to errors in the muscle forces estimation. The present study aims to propose a MSK model of the amputated gait that accounts for the interaction between the body and the ischial support for the estimation of the muscle forces of 13 subjects with unilateral transfemoral amputation. Contrary to previous studies on the amputated gait which ignored the interaction with the ischial support, here the contact on the ischial support was included in the external loads acting on the pelvis. Results show that including the ischial support induced an increase in the activity of the main abductor muscles while adductor muscles' activity was reduced. These results suggest that neglecting the interaction with the ischial support leads to erroneous muscle forces distribution considering the gait of people with transfemoral amputation. Although subjects with various bone geometries, particularly femur lengths, were included in the study, similar results were obtained for all subjects. Eventually, the estimation of muscle forces from MSK models could be used in combination with finite element models, to provide quantitative data for the design of prosthetic sockets.

Word count: 244

1. Introduction

A prosthetic socket must ensure the transmission of mechanical loads from the prosthesis to the skeleton of the person, and vice-versa. For people with transfemoral amputation, this is achieved by tightening the socket and, most of the time, by the inclusion of an ischial support in the socket design [1]. The ischial support is meant to decrease the need for tightening. Consequently, a percentage of the body weight is transmitted through the ischium and the soft tissues in-between and this is confirmed by experimental data [2]. In the frontal plane, the ischial support of the socket acts as a “pelvic lever”, as called by Radcliffe, and prevents the pelvis from dropping towards the contralateral side. At the same time, the mechanical action of the abductors maintains the pelvis horizontal [3]. Thus, the more abductor muscle forces the less the ischial support reaction force. Besides, the support on the ischial tuberosities is known to be harmful to many people with transfemoral amputation so pain and discomfort should be reduced if less weight is applied on the ischial support of the socket [4–6]. Finite Element Models (FEM) have been proposed previously as tools that help the prosthetists to design the socket and eventually limit the discomfort in the socket as much as possible [7–14]. Yet, current FEM are limited due to, among other things, boundary conditions that are poorly representative of the reality [15]. To improve the fidelity of FEM, some authors have proposed to include the muscle forces in FEM [16].

Muscle activation can be assessed experimentally, non-invasively, using surface ElectroMyoGram (EMG). However, EMG may only be used for superficial muscles and the conversion of the EMG signals into muscle forces can be tedious due to measurement artefacts, cross-talks and signal noises [17,18]. Furthermore, the positioning of EMG in the socket is challenging. Thus, a popular alternative is MusculoSkeletal (MSK) modelling [19–21].

Yet, MSK models of the amputated gait are limited [22]. In fact, to the authors’ knowledge, previous MSK models have systematically neglected the weight-bearing function of the ischial support of the socket. Some discrepancies have been observed between EMG and predicted muscle forces [23]. These could be attributed to the absence of the ischial support of the socket in the modelling procedure. Likewise, experimental studies concluded that ischial support is a major weight-bearing surface when measuring the pressure at the residual limb interface [24]. Neglecting the ischial support, in particular, in the frontal plane, may lead to non-negligible errors.

This study aims to design a MSK model of the amputated gait for individuals with unilateral transfemoral amputation that accounts for the contact force with the prosthetic socket on the ischial support contrary to previous studies on the amputated gait which ignored the interaction with the ischial support, here the contact on the ischial support was included in the external loads acting on the pelvis. The model was based on the rationale that people with transfemoral amputation sit in their sockets like on a bike seat [25]. By doing so, a portion of the body weight is transmitted by a vertical force applied by the ischial support to the ischium, through soft tissues. The author hypothesised that adding the force applied on the ischial support will mainly lead to a decrease in the abductor muscles' forces [26,27]. This model was used to quantify the impact of the load transmitted by the ischial support on the estimated muscle forces of 13 subjects.

2. Materials and methods

2.1. Participants

Power analysis was used to define the number of subjects with a type I error or 5 % and a type II error of 15 %[28]. To detect a difference of one standard deviation on the peak net hip abduction moment, at least 12 subjects are thus required. In total, thirteen subjects with unilateral transfemoral or Gritti amputations, with daily usage of their prosthesis, volunteered for this study (Table 1.). The analysis procedure of this study was approved by the *Comité de Protection des Personnes* (CPP NX06036).

Table 1 around here

2.2. Experimental acquisitions

Subjects were equipped with 55 optoelectronic markers on the lower limb [29] for the quantitative gait analysis that was performed according to the ISB recommendations [30,31]. Once the calibration in the static standing posture was performed, kinematic data were collected employing an optoelectronic motion analysis system (Vicon Nexus 2, Oxford Metrics Ltd, UK) with 13 cameras set at a frequency of 100 Hz. Ground reaction forces were simultaneously recorded with four force plates (AMTI Advanced Mechanical Technology, Inc, Massachusetts, USA) built-in level with the floor. Subjects walked at a self-selected speed and only complete strides (i.e., two successive foot contacts on the force plates) were saved. Markers trajectories and force plate data were low-pass filtered with a zero-phase 4th order Butterworth filter with a cut-off frequency of 5 Hz. A pair of X-rays was then acquired (EOS, EOS-Imaging, France) without removing the optoelectronic markers in the standard standing posture [32].

2.3. MSK modelling

2.3.1. *Generic geometries*

A generic MSK model including the pelvis and the femur, and the muscles detailed in **Erreur ! Source du renvoi introuvable.** was defined. Muscles’ respective origins, insertions and path points were defined according to the model proposed in the literature [33] except that all muscles’ path points were fixed to either the pelvis [34] or the femur [35].

2.3.2. *Personalisation*

Bi-planar X-rays were used to reconstruct the subject-specific geometries of the pelvis and the contralateral femur according to previous studies [36,37]. The femur was symmetrised to define the geometry of the ipsilateral femur before amputation which was then manually positioned to fit the X-rays. A kriging transformation was defined to compute the subject-specific positions of the muscles' points [38]. The surface mesh nodes of the generic and subject-specific geometries of the bones were the control points of the transformation in the source and the target domains respectively. The transformation was applied to the muscles' points to scale the muscles to the subject-specific geometries. The personalisation process was finalised by manually cutting the ipsilateral femur surface mesh at the level of the amputation computed from the bi-planar X-rays. All muscle' insertions below the amputation level were fixed at the distal end of the residual femur. Further details regarding the personalisation of the MSK models and the computation of the lever arms are provided in Supplementary Materials A.

2.3.3. *Muscle forces*

The subject-specific MSK models were computed from the personalised geometries using Matlab (The MathWorks, Inc., Matlab, California, USA). A ball-and-socket joint was used to model the hip joint. Hip joint angles during the gait cycle were retrieved from the motion capture data. The intersegmental hip moments expressed at the femoral head centre in the pelvis reference frame were computed from an inverse dynamic analysis. A static optimisation was used to estimate the muscle forces [39]. The algorithm was adapted to the amputated gait by decomposing the intersegmental hip moments ($M^{C}_{Residual\ limb \rightarrow pelvis}$) of the amputated side into the moment due to the discrete muscle forces ($F_{muscles}$), the moment due to the soft tissue contact loads at the ischial support level ($F_{soft\ tissues}$), and the moment due to the femur contact loads (F_{femur}) (Figure 1). Considering the hip joint as a ball-and-socket joint, the moment due to F_{femur} is null. The contribution of $F_{soft\ tissues}$ to the intersegmental hip moment was expressed as a reduction of the abduction component of the intersegmental moment [3,25]. Three levels of reduction, *k*, due to the moment of the soft tissues contact force at the ischial support were considered (0 %, 50 % and 100 % reduction).

All these hypotheses led to the following system of equations:

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

$$(2) \qquad \begin{cases} \begin{pmatrix} r_{abd1} & \cdots & r_{abdn} \\ r_{rot1} & \cdots & r_{rotn} \\ r_{flex1} & \cdots & r_{flexn} \end{pmatrix} \times x = \begin{bmatrix} (1-k) * M_{abd} \\ M_{rot} \\ M_{flex} \end{bmatrix}, k \in [0\,; 0.5\,; 1] \\ 0 \leq x \leq 2F^{max} \end{cases}$$

With J, the cost function to minimise, x the vector of all muscle forces, F^i the force of the ith muscle, F^i_{max} the maximal isometric force of the ith muscle assessed from literature data [33]. The maximal isometric force was multiplied by two as proposed by [40]. F_{max}, a n-by-1 vector, contains the maximal isometric forces of all muscles. The kinematic analysis and the 3D models of the bones were used to compute r_{abd}^i, r_{rot}^i and r_{flex}^i, the lever arms of the ith muscle relative to the hip centre respectively in abduction/adduction, internal/external rotation and flexion/extension [41]. M_{abd}, M_{rot} and M_{flex}, are the intersegmental hip moment components in abduction/adduction, internal/external rotation and flexion/extension respectively and for the total number of muscles. The moment due to the muscle forces will be referred to as the hip muscular moment in the following. When k was equal to 1, 100 % of the intersegmental hip abduction moment was compensated by the moment induced by the contact force at the ischial support level. On the contrary, when k was null, there was no reduction of the intersegmental hip moment in the frontal plane which is thus equal to the hip muscular moment. This meant that no weight was applied to the ischial support. The optimisation was performed using the *fmincon* with initial forces set to zero built-in MATLAB function.

FIGURE 1 around here

3. Results

3.1. Personalised models

The personalised MSK models of the 13 volunteers are shown in Figure 2.

FIGURE 2 around here

3.2. Intersegmental hip moment

Mean and normality corridor of the intersegmental hip moment of the amputated side at the hip joint centre, are presented in Figure 3. Mean peak values from the current studies are 0.42 N.m.kg^{-1}, 0.16 N.m.kg^{-1} and, 0.71 N.m.kg^{-1} for the moments in the frontal, transverse and sagittal planes respectively.

FIGURE 3 around here

3.3. Muscle forces

Validation of the muscle lever arms is in Supplementary Materials B. Muscle forces were estimated in three scenarios corresponding to the values of factor k (see equation 2). Changing the value of k had a net impact on the estimation of the muscle forces (Figure 4a, Figure 4b). The maximal variation of the mean muscle forces over the gait cycle of all subjects is detailed in Table 2. The force of the gluteus medius muscle, as the main abductor muscle, is the most affected by the change of k, with a mean decrease of 7.6 N.kg^{-1} from $k = 0$ to $k = 1$. Regarding adductor muscles, the forces of the adductor magnus and the biceps femoris were the two most impacted by the modelling of the prosthesis with a maximal mean increase of 1.8 N.kg^{-1} and 1.4 N.kg^{-1} of the muscle forces.

FIGURE 4a and 4b around here

Table 2 around here

4. Discussion

The objective of this study was to propose a new optimisation procedure for the gait of people with transfemoral amputation that could account for the force applied by the ischial support to the residual limb. This was achieved by designing a subject-specific musculoskeletal model of the amputated gait. In this study, it was considered that the external force applied by the ischial support compensates for a certain percentage of the net hip intersegmental abduction moment. Consistent with the authors' hypothesis, accounting for the force applied by the ischial support mainly impacted the abductor muscles acting on the hip joint, yet this was not the only conclusion of this study. The increase of the reduction factor also affected the main adductor muscles, adductor magnus and biceps femoris, for which the level of activation increased.

This study showed that the reduction of the abductor muscles' forces solely may not be sufficient to maintain the moment equilibrium if an important reduction occurs. On the other hand, these muscles also act on the extension of the hip. Once again, to maintain the moment equilibrium some flexor muscles such as the psoas or the rectus femoris were less used. Therefore, the reduction factor was critical to estimate the muscle forces. The results highlighted that the required muscle forces to walk are modified when subjects apply a different percentage of their body weight on the ischial support. This is directly and indirectly related to the subject's weight but also to the amputation level, the type of socket or other prosthetic devices. In other words, the reduction factor should be chosen carefully when performing MSK modelling of the amputated gait. People with Gritti amputations may have a reduction factor close to zero since they are less likely to use the ischial support that is often missing from their sockets. People with low mobility or weak muscle structures may have a reduction factor close to 1 since they will apply more weight to their ischial support. A possible assessment of this factor may be performed with measurement of the pressure distribution in the socket [42].

Results were compared to the literature data. Mean peak moment values estimated in the current study were 0.42 $N.m.kg^{-1}$, 0.16 $N.m.kg^{-1}$ and, 0.71 $N.m.kg^{-1}$ for the moments in the frontal, transverse and sagittal planes respectively. Previous studies presented moments comprised between 0.43 and 0.71 $N.m.kg^{-1}$ in the frontal plane [43,44], 0.14 and 0.18 $N.m.kg^{-1}$ in the transverse plane [43], and 0.30 and 1.80 $N.m.kg^{-1}$ in the sagittal plane [43–46]. Considering muscle forces estimation, the complete range of the force applied by the soft tissues on the ischial support, from a reduction of 0 % to 100 % of the intersegmental hip moments in the frontal plane, was investigated. The results with no reduction of the intersegmental hip moments in the frontal plane were compared to the literature data. A mean peak force at 25 % of the gait cycle comprised between 1 and 3 N.kg^{-1} was reported for the gluteus maximus, between 4 and 7 N.kg^{-1} for the gluteus medius, 0 and 1 N.kg^{-1} for the hamstrings, 0 and 2 for the iliopsoas, 1 and 3 N.kg^{-1} for the rectus femoris by Harandi et al. [19]. Mohamed also reported a force comprised between 0 and 2 N.kg^{-1} for the rectus femoris, 0 and 1 for the biceps femoris, and 0 and 1 N.kg^{-1} for the adductor magnus [47]. In the current study, the gluteus maximus force was comprised between 0 and 1 N.kg^{-1}, the gluteus medius force between 2 and 15 N.kg^{-1}, the biceps femoris force was below 1 N.kg^{-1}, the iliopsoas between 0 and 6 N.kg^{-1}, the rectus femoris between 0 and 3 N.kg^{-1}, and the adductor magnus force was below 1 N.kg^{-1}. The range of values was larger than those presented in literature studies probably due to the larger number of volunteers with various residual limb lengths but also by the saturation of the flexor muscle forces for some subjects. This highlights the need to personalise the muscle parameters towards the study of the amputated gait.

Some limitations of the current study may have influenced the results. Residual femur and muscles' insertion points may need a more accurate personalisation since the muscles' distal anchoring performed during the amputation has a significant impact on the muscle activations [21]. This could be investigated using MRI [48] or ultrasound imaging. However, such processes require further data acquisition and image processing which is time-consuming. The simplification of the muscles modelling must have also led to erroneous estimations of the lever arms which may explain some of the discrepancies with other literature studies. However, as shown in supplementary materials, most lever arms were consistent with literature data, except in some cases in the transverse plane [49]. Other errors in muscle forces may be explained by the optimisation process. For example, the maximal isometric forces were defined according to literature data of subjects without amputation [40]. Muscle atrophy due to the amputation was thus neglected in this study. Other parameters such as the tension-extension relationships of muscles and tendons were not considered because of the difficulties in identifying such parameters in subjects with amputation. Eventually, the model was not experimentally validated, yet results were compared to literature data of superficial EMG [50,51]. For all studies, the gluteus maximus main activation was during the beginning of the stance phase that is to say before 40 % of the gait cycle; the gluteus medius was activated during the entire stance phase; the tensor fascia latae was activated at the end of the stance phase and the beginning of the swing phase. Some discrepancies were noticed for the rectus femoris and the gracile which were activated sooner in the current study compared to the literature. On the contrary, the adductor magnus and the biceps femoris were activated later in the current study compared to the literature. Discrepancies may be explained by the previously detailed inaccuracies in the modelling but also by errors in the EMG measurements. EMG data are tedious to acquire due to the difficulties in positioning the sensors on the muscles in the socket, intramuscular crosstalk and low-intensity signals on deep muscles. As a result, literature data provide a wide range of muscle activations which makes the interpretation difficult [50,51].

Future improvements of the model would include adding the impact of the ischial support on the other components of the intersegmental hip moment particularly in the sagittal plane and the experimental validation of the come of the muscle forces using superficial EMG. Experimental measurements with force sensors in the socket could also be performed to assess the external loads applied on the ischial support. Other acquisitions using ultrasound imaging systems could be used to personalise the muscles' parameters such as the maximal isometric force using the physical cross-sectional area of muscles or the muscles' insertions and origins.

Thus far the proposed results highlight the fact that the ischial support of the socket should not be neglected to compute muscle forces. By relying too much on ischial support, people with transfemoral amputation may reduce the abductor muscles’ activity which may amplify the atrophy phenomenon occurring after the amputation. This could also lead to more pain and discomfort at the ischium level. Therefore, when the socket fit fails to provide comfortable support, one may consider orientating the rehabilitation process towards the strengthening of the abductor muscles. In addition, prosthetists could make the most of such data to design the subject-specific socket using FEM to estimate the pressure at the interface of the residual limb. Indeed, a recent study has shown that the hip joint modelling, i.e., bones and muscles acting on the joint degrees of freedom, was necessary for the estimation of interface pressure [52]. Considering advances in reduced order models and artificial intelligence based technics, combined MSK and FEM could bring new insights for the prosthetist practice.

5. Conclusion

The use of a prosthesis may be painful for people with transfemoral amputation especially at the level of the ischial support where a non-negligible part of the body weight is applied. One potential avenue to diminish the support on the ischial tuberosities is to increase the contribution of the abductor muscles. Eventually, the socket could be designed considering the force that the abductor muscles, and more particularly, the gluteus medius is able to develop. To analyse this contribution, MSK models can be used. This method allows to also account for the contribution of muscles and other external forces applied to the residual limb such as the one due to the stance on the ischial support. This study showed the important impact of the ischial support on the estimated muscle forces, particularly considering the abductor muscles acting on the hip joint. With further improvement and after experimental validation of these results, such models could be used clinically to assess the muscular activity during gait of people with transfemoral prothesis and in finite element analysis to compute the pressure distribution applied by the socket on the residual limb.

6. Conflict of interest

The authors certify that no conflict of interest is raised by this work.

7. References

1. Kapp SL. Transfemoral Socket Design and Suspension Options. *Phys Med Rehabil Clin N Am*. 200AD;11(3):569-584.

2. Appoldt FA, Bennett L. A preliminary report on dynamic socket pressures. *Bull Prosthet Res*. 1967;10(8):20-55.

3. Radcliffe CW. Functional considerations in the fitting of above-knee prostheses. *Artif Limbs*. 1955;2(1):35-60. doi:10.1017/CBO9781107415324.004

4. Lee VSP, Solomonidis SE, Spence WD. Stump-socket interface pressure as an aid to socket design in prostheses for trans-femoral amputees—a preliminary study. *Proc Inst Mech Eng H*. 1997;211(2):167-180. doi:10.1243/0954411971534287

5. Kahle JT, Highsmith MJ. Transfemoral sockets with vacuum-assisted suspension comparison of hip kinematics, socket position, contact pressure, and preference: Ischial containment versus brimless. *J Rehabil Res Dev*. 2014;50(9):1241-1252. doi:10.1682/jrrd.2013.01.0003

6. Kahle J, Miro RM, Ho LT, et al. The effect of the transfemoral prosthetic socket interface designs on skeletal motion and socket comfort: A randomized clinical trial. *Prosthet Orthot Int*. 2020;44(3):145-154. doi:10.1177/0309364620913459

7. Meng Z, Wong DWC, Zhang M, Leung AKL. Analysis of compression/release stabilized transfemoral prosthetic socket by finite element modelling method. *Med Eng Phys*. 2020;83:123-129. doi:10.1016/j.medengphy.2020.05.007

8. Submitted AD, The TO, Of S. Evaluation of Trans-femoral Prosthesis Function Using Finite Element Analysis. 2017;(September).

9. Zhang L, Zhu M, Shen L, Zheng F. Finite element analysis of the contact interface between trans-femoral stump and prosthetic socket. In: *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*. ; 2013:1270-1273. doi:10.1109/EMBC.2013.6609739

10. Goh JCH, Lee PVS, Toh SL, Ooi CK. Development of an integrated CAD-FEA process for below-knee prosthetic sockets. *Clinical Biomechanics*. 2005;20(6):623-629. doi:10.1016/j.clinbiomech.2005.02.005

11. Abu Osman N a, Spence WD, Solomonidis SE, Paul JP, Weir a M. The patellar tendon bar! Is it a necessary feature? *Med Eng Phys*. 2010;32(7):760-765. doi:10.1016/j.medengphy.2010.04.020

12. Lacroix D, Patiño JFR. Finite element analysis of donning procedure of a prosthetic transfemoral socket. *Ann Biomed Eng*. Published online 2011. doi:10.1007/s10439-011-0389-z

13. Wagner DW, Divringi K, Ozcan C, Grujicic M, Pandurangan B, Grujicic A. Combined musculoskeletal dynamics/structural finite element analysis of femur physiological loads during walking. *Multidiscipline Modeling in Materials and Structures*. 2010;6(4):417-437. doi:10.1108/15736101011095118

14. Lin CC, Chang CH, Wu CL, Chung KC, Liao IC. Effects of liner stiffness for trans-tibial prosthesis: A finite element contact model. *Med Eng Phys*. 2004;26(1):1-9. doi:10.1016/S1350-4533(03)00127-9

15. Taylor M, Prendergast PJ. Four decades of finite element analysis of orthopaedic devices: Where are we now and what are the opportunities? *J Biomech*. 2015;48(5):767-778. doi:10.1016/j.jbiomech.2014.12.019

16. Fougeron N, Rohan P yves, Rose J loïc, Bonnet X, Pillet H. Finite element analysis of the stump-ischial containment socket interaction: a technical note. *Med Eng Phys*. 2022;105.

17. Farina D, Merletti R, Indino B, Nazzaro M, Pozzo M. Surface EMG crosstalk between knee extensor muscles: Experimental and model results. *Muscle Nerve*. 2002;26(5):681-695. doi:10.1002/mus.10256

18. Disselhorst-Klug C, Schmitz-Rode T, Rau G. Surface electromyography and muscle force: Limits in sEMG-force relationship and new approaches for applications. *Clinical Biomechanics*. 2009;24(3):225-235. doi:10.1016/j.clinbiomech.2008.08.003

19. Harandi VJ, Ackland DC, Haddara R, et al. Gait compensatory mechanisms in unilateral transfemoral amputees. *Med Eng Phys*. 2020;77:95-106. doi:10.1016/j.medengphy.2019.11.006

20. Bae TS, Choi K, Hong D, Mun M. Dynamic analysis of above-knee amputee gait. *Clinical Biomechanics*. 2007;22(5):557-566. doi:10.1016/j.clinbiomech.2006.12.009

21. Ranz EC, Wilken JM, Gajewski DA, Neptune RR. The influence of limb alignment and transfemoral amputation technique on muscle capacity during gait. *Comput Methods Biomech Biomed Engin*. 2017;20(11):1167-1174. doi:10.1080/10255842.2017.1340461

22. Anderson FC, Pandy MG. A dynamic optimization solution for vertical jumping in three dimensions. *Comput Methods Biomech Biomed Engin*. 1999;2(3):201-231. doi:10.1080/10255849908907988

23. Toderita D, Henson DP, Klemt C, Ding Z, Bull AMJ. An Anatomical Atlas-Based Scaling Study for Quantifying Muscle and Hip Joint Contact Forces in above and Through-Knee Amputees Using Validated Musculoskeletal Modelling. *IEEE Trans Biomed Eng*. 2021;68(11):3447-3456. doi:10.1109/TBME.2021.3075041

24. Colombo G, Comotti C, Redaelli DF, Regazzoni D, Rizzi C, Vitali A. A method to improve prosthesis leg design based on pressure analysis at the socket-residual limb interface. *Proceedings of the ASME Design Engineering Technical Conference*. 2016;1A-2016:1-8. doi:10.1115/DETC2016-60131

25. Pritham CH. Biomechanics and shape of the above-knee socket considered in light of the ischial containment concept. Published online 1990:9-21.

26. Radcliffe CW. Functional considerations in the fitting of above-knee prostheses. *Artif Limbs*. 1955;2(1):35-60. doi:10.1017/CBO9781107415324.004

27. Radcliffe CW. Above-knee prosthetics. Published online 1977:146-160.

28. Hansen A, Nickel E, Medvec J, Brielmaier S, Pike A, Weber M. Effects of a flat prosthetic foot rocker section on balance and mobility. *J Rehabil Res Dev*. 2014;51(1):137-148. doi:10.1682/JRRD.2012.12.0229

29. Pillet H, Drevelle X, Bonnet X, et al. APSIC: Training and fitting amputees during situations of daily living. *Irbm*. 2014;35(2):60-65. doi:10.1016/j.irbm.2014.02.005

30. Wu G, Cavanagh PR. *ISB RECOMMENDATIONS FOR STANDARDIZATION IN THE REPORTING OF KINEMATIC DATA*. Vol 28.; 1995.

31. Wu G. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *J Biomech*. 2002;35:543-548.

32. Facione J, Villa C, Bonnet X, et al. Spinopelvic sagittal alignment of patients with transfemoral amputation. *European Spine Journal*. 2019;28(9):1920-1928. doi:10.1007/s00586-019-06017-x

33. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: Open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng*. 2007;54(11):1940-1950. doi:10.1109/TBME.2007.901024

34. Dumas R, Robert T, Cheze L, Verriest JP. Thorax and abdomen body segment inertial parameters adjusted from McConville et al. and Young et al. *Int Biomech*. 2015;2(1):113-118. doi:10.1080/23335432.2015.1112244

35. Cappozzo a, Catani F, Della Croce U, Leardini a. Position and orietnation in space of bones during movement. *Clin Biomech*. 1995;10(4):171-178.

36. Mitton D, DeschêNes S, Laporte S, et al. 3D reconstruction of the pelvis from bi-planar radiography. *Comput Methods Biomech Biomed Engin*. 2006;9(1):1-5. doi:10.1080/10255840500521786

37. Chaibi Y, Cresson T, Aubert B, et al. Fast 3D reconstruction of the lower limb using a parametric model and statistical inferences and clinical measurements calculation from biplanar X-rays.

Comput Methods Biomech Biomed Engin. 2012;15(5):457-466. doi:10.1080/10255842.2010.540758

38. Trochu F. A contouring program based on dual kriging interpolation. *Eng Comput*. 1993;9(3):160-177. doi:10.1007/BF01206346

39. Crowninshield RD, Brand RA. A physiologically based criterion of muscle force prediction in locomotion. *J Biomech*. 1981;14(11):793-801. doi:10.1016/0021-9290(81)90035-X

40. Arnold EM, Hamner SR, Seth A, Millard M, Delp SL. How muscle fiber lengths and velocities affect muscle force generation as humans walk and run at different speeds. *Journal of Experimental Biology*. 2013;216(11):2150-2160. doi:10.1242/jeb.075697

41. Zajac FE, Gordon ME. Determining Muscle's Force and Action in Multi-Articular Movement. *Exercicse and Sport Sciences Reviews*. 1989;17(1):187-230.

42. Laszczak P, McGrath M, Tang J, et al. A pressure and shear sensor system for stress measurement at lower limb residuum/socket interface. *Med Eng Phys*. 2016;38(7):695-700. doi:10.1016/j.medengphy.2016.04.007

43. Frossard L, Cheze L, Dumas R. Dynamic input to determine hip joint moments, power and work on the prosthetic limb of transfemoral amputees: Ground reaction vs knee reaction. *Prosthet Orthot Int*. 2011;35(2):140-149. doi:10.1177/0309364611409002

44. Heitzmann DWW, Leboucher J, Block J, et al. The influence of hip muscle strength on gait in individuals with a unilateral transfemoral amputation. *PLoS One*. 2020;15(9):e0238093. doi:10.1371/journal.pone.0238093

45. Sjödahl C, Jarnlo GB, Söderberg B, Persson BM. Kinematic and kinetic gait analysis in the sagittal plane of trans-femoral amputees before and after special gait re-education. *Prosthet Orthot Int*. 2002;26(2):101-112. doi:10.1080/03093640208726632

46. Segal AD, Orendurff MS, Klute GK, et al. Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg® and Mauch SNS® prosthetic knees. *J Rehabil Res Dev*. 2006;43(7):857-870. doi:10.1682/JRRD.2005.09.0147

47. Mohamed A. *Modeling and Simulation of Transfemoral Amputee Gait*. University of New Brunswick; 2018.

48. Suzuki Y. Dynamic optimization of transfemoral prosthesis during swing phase with residual limb model. *Prosthet Orthot Int*. 2010;34(4):428-438. doi:10.3109/03093646.2010.484829

49. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: Open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng*. 2007;54(11):1940-1950. doi:10.1109/TBME.2007.901024

50. Jaegers SMHJ, Arendzen JH, De Jongh HJ. An electromyographic study of the hip muscles of transfemoral amputees in walking. *Clin Orthop Relat Res*. 1996;(328):119-128. doi:10.1097/00003086-199607000-00020

51. Wentink EC, Beijen SI, Hermens HJ, Rietman JS, Veltink PH. Intention detection of gait initiation using EMG and kinematic data. *Gait Posture*. 2013;37(2):223-228. doi:10.1016/j.gaitpost.2012.07.013

52. Fougeron N, Rohan P yves, Rose J loïc, Bonnet X, Pillet H. Finite element analysis of the stump-ischial containment socket interaction: a technical note. *Med Eng Phys*. 2022;105. doi:https://doi.org/10.1016/j.medengphy.2022.103829

8. List of figures

Figure 1: Modelling approach proposed for the static optimisation algorithm. a) The external loads applied to the pelvis segment (T) are expressed at the centre of the femoral head (C). b) The external loads included the forces of the muscles (FMuscles) applied at the insertions points of each muscle, the force of the femur (FFemur) at the centre of the femoral head (C) and the force of the soft tissues at the ischial support level(FSoft Tissues) expressed here at a contact point located on the ischial plateau support of the socket (P).

Figure 2: Subject-specific bones and muscles geometries for the 13 subjects of the database.

Figure 3: Mean intersegmental hip moment expressed in the pelvis reference frame in the three directions. Normality corridors were computed with the mean values plus/minus the standard deviation.

Figure 4a: Mean muscle forces of the first half of the muscles estimated with MSK modelling during a complete walk cycle for all subjects when considering three levels of reduction of the net hip abduction moment: 0% (blue), 50 % (red) and 100 % (yellow). Normality corridors correspond to the mean values plus/minus one standard deviation. The horizontal red line account for the maximal isometric force input for each muscle.

Figure 4b: Mean muscle forces of the second half of the muscles estimated with MSK modelling during a complete walk cycle for all subjects when considering three levels of reduction of the net hip abduction moment: 0% (blue), 50 % (red) and 100 % (yellow). Normality corridors correspond to the mean values plus/minus one standard deviation. The horizontal red line account for the maximal isometric force input for each muscle.

9. List of tables

Table 1: Subjects and prothesis features.

Table 2: Maximal changes in mean muscle forces of all subjects during the complete gait cycle among the three values of intersegmental moment reduction, k varying from 0 to 1.