



Seeking for a better Human-Prosthesis energetic gait efficiency by quantifying both propulsion power and instability control

Helene Pillet, Xavier Bonnet, Amandine Boos, Lucas Sedran, Bruno Watier

► To cite this version:

Helene Pillet, Xavier Bonnet, Amandine Boos, Lucas Sedran, Bruno Watier. Seeking for a better Human-Prosthesis energetic gait efficiency by quantifying both propulsion power and instability control. 2022 IEEE-RAS 21st International Conference on Humanoid Robots (Humanoids), Nov 2022, Ginowan, Japan. pp.566-571, 10.1109/Humanoids53995.2022.10000069 . hal-04239299

HAL Id: hal-04239299

<https://laas.hal.science/hal-04239299>

Submitted on 12 Oct 2023

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

Seeking for a better Human-Prosthesis energetic gait efficiency by quantifying both propulsion power and instability control *

Helene Pillet¹, Xavier Bonnet¹, Amandine Boos¹, Lucas Sedran¹ and Bruno Watier²

Abstract—The present study aims at quantifying propulsion and dynamic balance through biomechanical parameters issued from theoretical modeling and analysis of locomotion during the gait of people using prosthetic devices. An experimental protocol combined motion capture and oxygen consumption quantification during gait on a treadmill. The mechanical work produced and dissipated by the lower limbs and the evolution of a biomechanical indicator of balance were used and the estimation of the metabolic cost of walking was made from oxygen consumption. To test the relevance of the chosen parameters, the experiments were performed on six able-bodied volunteers successively equipped with two prosthetic ankle-feet (elastic vs rigid) mounted on a femoral prosthetic simulator. For each participant, the parameters were computed and compared in three configurations: i/ without prosthesis, ii/ with rigid prosthetic ankle-foot iii/ with elastic prosthetic ankle-foot. The results put in evidence an increase of energy consumption in both prosthetic configurations compared to the configuration without prosthesis. However, no differences could be observed between the elastic and rigid prosthetic configurations. The analysis of mechanical work performed by each lower limb, which confirmed the energy delivered by the elastic foot during the propulsion, did not explain by its own this discrepancy. The maintenance of balance that seems to be more challenging during the double support in the elastic configuration could be involved in this counter-intuitive result. Finally, this preliminary study shows the importance to consider simultaneously propulsion and balance objectives during gait as they must both require muscular actions involved in the production of energy by the prosthesis user.

I. INTRODUCTION

Poor human-prosthesis adaptation has been identified as a key factor of the relative inefficiency of robotic lower limb prostheses to restore a normal metabolic cost of transport of individuals with lower limb amputation [1]. Indeed, if considered as an isolated parameter, ankle push-off is clearly related to the energetic efficiency of walking in humans [2]. However, powered prosthetic ankles failed in restoring both normal energetic consumption and preferred gait speed of people with transtibial amputation [3]. A possible explanation lies in the independent control of the prosthesis and the human musculo-skeletal system. In normal locomotion, the joints are actuated by muscles including bi-articular ones that ensure consistent actions under the control of the central nervous system. While wearing a prosthesis,

both prosthetic and residual musculo-skeletal system are not directly connected anymore, which can hinder their correct interaction. Considering people with transfemoral amputation, this should be emphasized as they have to use simultaneously a prosthetic knee and a prosthetic ankle-foot, which behavior are generally independent from each other. On another hand, dynamic balance during locomotion is also known as challenging due to the inherent instability of biped walking. Besides, the control of instability is intuitively involved in the overall energetic cost of walking as it requires muscle activations to prevent falling. Again, the challenge is all the more important for persons wearing prosthetic devices that are known to have more asymmetrical and perturbed kinematics and dynamics. However, very few studies have sought to evaluate simultaneously ‘balance-related effort’ and push-off work control [4]–[5].

A. State of the art

The study of the energetics of gait has been the topic of numerous biomechanical studies. To this aim, the estimation of the metabolic energy through oxygen consumption quantification is considered as a gold standard to analyze the effect of different configurations for a given individual at a given time [6] even if the results are known to be sensitive to the well-known biases associated to great interindividual variations, influence of nutrition and fatigue. To better understand the link with muscle functions, several teams have tried to link the metabolic energy consumed by the individuals to mechanical works performed by the joints [7]–[11]. The definition of mechanical work that should be correlated to metabolic energy is also subject to debate. The definition depends on the underlying modeling, of which the complexity can vary. Work of the Individual Limb Method (WILM) [12] seems a good compromise being simple to assess and closely related to other methods such as summed-joint-power classically considered as a standard in biomechanics [13]. To date, more complex models such as complete musculo-skeletal modeling of the body have not led to significantly more accurate estimation of the metabolic cost [14]. As concerns gait instability assessment, several indicators have also been proposed [15]. Among them, the variation of the angular momentum at the body center of mass (BCoM) during the motion appears as one of the most promising due to its ability to link the rotations of the segments to the external mechanical action at the center of mass [16]. However, both its determination and interpretability can be complex. In a recent work, we propose to use a novel mechanical indicator of the instability of gait that partly solve

*Research supported by the Fondation de l’Avenir (grant number AP-RM-20-001), DGA-RAPID AAFAP and ANR Hobis (grant number 18CE27-0010) and COBOT (grant number 18CE10 – 0003).

¹H. Pillet, X. Bonnet, A. Boos and L. Sedran are with Arts et Metiers Sciences and Technologies Institut de Biomecanique Humaine Georges Charpak, Paris, France. helene.pillet@ensam.eu

²B. Watier is with LAAS-CNRS, Toulouse, France bruno.watier@laas.fr

these issues [17], [18]. It is based on a distance computed between the Minimal Moment Axis of external mechanical actions to the Body Center of Mass (MMA/BCoM). To our knowledge, no study has analyzed for the same subjects, the quantification of mechanical work, of gait instability and of the consumed metabolic energy in a population of people with transfemoral amputation. One reason of this lack is the difficulty to recruit such volunteers. To solve the issue of recruitment, some authors have proposed to use prosthetic simulators that can be worn by non-amputee persons [19], [20]. They have been widely used for prosthetic ankle-foot evaluation but never for knee and ankle-foot replacement.

B. Contributions

In the present work, we propose to combine both complementary approaches of mechanical work quantification and gait instability assessment and to analyze then in regards of the quantification of the metabolic energy. We therefore proposed to determine ILM mechanical work (WILM) together with the MMA/BCoM indicator described in the above section. Both parameters determination relies on the computation of body center of mass kinematics and external mechanical actions during gait. Metabolic energy was assessed independently from oxygen consumption uptake during walking.

In order to test the relevance and sensitivity of the parameters, we designed an original experimental framework involving volunteers equipped with prosthesis simulator including knee and ankle-foot joints. Three conditions could be compared without and with the simulator equipped with two different prosthetic foot. Finally, the implications of the experimental results for prosthesis design and control have been extracted.

II. THEORETICAL BACKGROUND

A. Mechanical work and energetic considerations

There exist different ways of quantifying mechanical power and work performed by the joints when walking. The kinetic energy theorem links the variation of kinetic energy E_{kin} of a system (here the entire body) relatively to a referent frame (here R_o) to the sum of the powers of external and internal mechanical actions (P_{ext} and P_{int}) (1).

$$\frac{dE_{kin}(body/R_o)}{dt} = P_{ext} + P_{int} \quad (1)$$

Considering the entire body as the mechanical system of interest and modeling the body segments as articulated rigid bodies, the sources of internal power originate from the mechanical actions in the anatomical joints, which themselves result from the muscles mechanical actions passing through the joints. Therefore, at steady state, for example during a gait cycle the metabolic energy that is consumed for the functioning of muscles generating joint motions should theoretically be inferred from the computation of mechanical work performed inside the body [21]. Mechanical work quantification over a given period of time can be done by integrating mechanical power. However, as this computation

stands on an algebraic sum of powers, it implicitly assumes that the mechanical energy produced in a joint can be absorbed by another one whether it is in the same limb or between limbs [9]. If this is correct in a mechanical system, the physiologic reality could be different as no evidence that such types of transfer exist between muscles [9] and the current level of knowledge does not allow to integrate refined modeling of muscle functioning in an analysis including the whole body. Taking a more pragmatic approach, some authors have proposed to sum the joint power no more in an algebraic sense but by using absolute values either considering individual joints [9], [22], [23] or individual limbs [24]. In the individual limbs paradigm, the underlying modeling of gait is a succession of pendulum-like phases where power is mainly generated or dissipated during the step-to-step transition [25], [26]. The internal mechanical power performed by the body can therefore be computed from the scalar product of the ground reaction forces (equal to the internal force in the pendulum) and the velocity of the center of mass. The comparison of this estimation of the internal power with the one made from the multibody modeling has already been performed from experimental data [27], [28] and from a theoretical approach [26] for level walking.

B. Control of the instability during locomotion

In addition to provide the necessary energy to achieve a given displacement of the body center of mass, the human musculo-skeletal system must be controlled in a way to avoid falling. In the case of generic biped walking, the external mechanical actions that act on the system are gravity and ground contact forces, knowing that contacts are regularly broken (toe off) and recovered (heel strike) [29]. In mechanics, stability refers to a given equilibrium position of the system (where the movement parameters remain constant) and can be defined as the ability of the system to recover from a perturbation around that equilibrium position. Thus, walking is not a succession of equilibrium positions and cannot be considered as stable from this point of view [29]. From such mechanical considerations, it does not appear straightforward to relate mechanical modeling to comprehensive parameters of balance capacity. The dynamic laws directly relate the acceleration of the center of mass of the system to the external resulting forces acting on it and the variations of the angular momentum at the body center of mass (which includes the effect of inertia and the rotational and translational accelerations of the individual body segments) to the resulting moment from mechanical actions at the center of mass. A variety of parameters have been derivated from this type of analysis involving more or less complex modeling of the human body used in both biomechanics and robotics among which Zero Moment Point (ZMP)/Center of Pressure (COP) is the most popular [30]. Its relative position to the support basis has been used as an objective for robot control and for biomechanical evaluation of gait stability. However, this criterion has revealed limited and not applicable in several conditions because it assumes that the contact points are on

the same horizontal surface and internal muscles or motor actions alleviate immediate fallings. Therefore, alternative parameters involving the kinematics of the center of mass (Margin of Stability [31], Foot Rotation Index [32], ...) have been developed. At the same time, a thorough analysis of the amplitude of the whole-body angular momentum at the center of mass during human gait [33]–[34] showed that it is controlled during steady-state gait and increases in impaired populations [33], [34]. The variations of the whole-body angular momentum can be also captured from the ground reaction forces and moments. This is the basis for the definition of the Centroidal Moment Point developed by Herr and Popovic [33] or the Virtual Pivot Point proposed by Maus [35]. However, both consider a simplified representation of the mechanical actions of contact between the feet and the ground by neglecting in particular the free moment which remains at the center of pressure around the vertical axis. On the contrary, our team proposed a parameter called MMA/BCoM index which consists in the computation of the distance between the center of mass (BCoM) of the system and the minimum moment axis (MMA) of the external mechanical actions of the ground on the system [17],[18]. Indeed, the theory of mechanical actions modeling shows that there is an axis called the central axis or minimum moment axis along which the ground reaction moment is minimal [33]. The distance of this axis relatively to the center of mass $d_{BCoM/MMA}$ is therefore directly related to the moment of external actions acting at the center of mass \mathbf{M}_{BCoM} and the ground reaction force \mathbf{F} , with the advantage of being easy to understand (2). The greater the distance, the greater the angular acceleration of the body around the center of mass is.

$$d_{BCoM/MMA} = \frac{\|\mathbf{F} \times \mathbf{M}_{BCoM}\|}{\|\mathbf{F}^2\|} \quad (2)$$

III. EXPERIMENTAL ASSESSMENT

A. Prosthetic configurations

Starting from the statements that (i) most of the power is delivered at the ankle joint during normal gait and (ii) larger increase of energy consumption has been observed for transfemoral amputee, we have chosen to compare two femoral prostheses in which only the prosthetic ankle-foot varied. The prosthetic knee was a magneto-rheologic microprocessor-controlled knee (*RheoTM*, Ossur). The two prosthetic ankle-feet were a carbon-leaf elastic foot, referred thereafter as ESR (*DynatreckTM*, Proteor) and a rigid 3D printed foot, referred thereafter as RF Fig.1. The stiffness of the ESR foot was chosen according to the mass of each person in accordance with prosthetist recommendations. The rigid foot was designed to reproduce a similar evolution of the center of pressure in the tibial local frame for the RF as compared to the ESR [36].

B. Participants and prosthesis simulator

To avoid the difficulties associated to the inclusion of persons with amputation (recruitment, security), we proposed

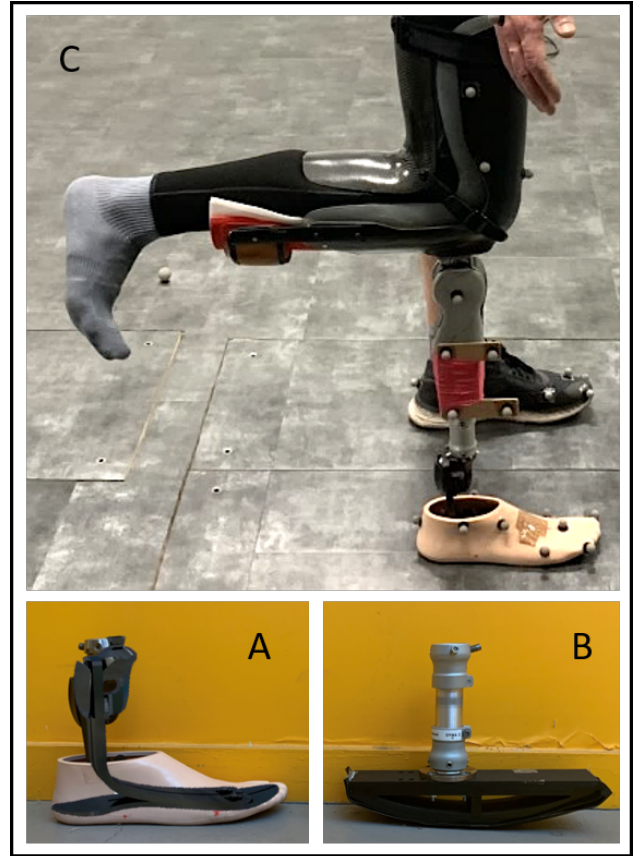


Fig. 1. Prosthetic feet used (A) Energy Storing and Release and (B) rigid and prosthesis simulator (C)

to use a prosthesis simulator that can be worn by non-amputee persons (Fig.1). It also allows to compare the two prosthetic configurations to the non-amputee configuration where the same individuals walked without the simulator. Six able-bodied volunteers (2 females and 4 males) participated in the study (mean age 31 years (std 8.6), mean height 174 cm (std 5.8), mean mass 68 kg (std 10.9). They were trained (at least one hour a day during one week) to walk with the simulator before the experiments.

C. Experimental protocols

The protocol (2020-A02369-30) was approved by ethical committee and each participant signed an informed consent. The protocol was divided in two sessions in which they successively underwent a quantitative gait analysis from motion capture system (ViconTM, Oxford Metrics®) and force plates acquisition (AMTI®) and an analysis of metabolic energy consumption using a respirometry system (K5TM, Cosmed®). During the first session, they were asked to walk on a 10 m flat walkway including three separate force plates (two 586*464cm and one 600*800 cm) in series. The contact reaction mechanical actions between each foot and the ground could therefore be acquired at 1000 Hz during three successive stance phases. In the same time, subjects were equipped with 58 reflective markers that were captured

synchronously by the optoelectronic system at 200 Hz. The subjects chose their own self-selected speed with the prosthesis simulator equipped with the rigid foot and reproduced it for all other configurations (elastic foot and without the simulator). In the second session, the same subjects walked in the same three configurations at the velocity chosen in the first session on a treadmill for 6 min per configuration. Before starting the experiment, the resting metabolic energy consumption was measured for 3 min for baseline while the subject stood quietly. Then, oxygen consumption was continuously recorded during walking.

D. Data acquisition and processing

From the first session, only the trials with three successive foot contacts on the force plates were considered. Force plates data and markers' trajectories were filtered using a 4th order low-pass Butterworth filter with a 15 Hz cut-off frequency. A 15-segment hybrid inertial model defined according to [37] was used to obtain body segmental inertial parameters, and the BCoM instantaneous position was computed as the weighted sum of the segments' centers of mass. Then, the mechanical power of each individual limb was assessed by the dot product of the velocity of the body center of mass (by derivation of its position) and the resulting ground reaction forces on the considered lower limb. Positive and negative works (WILM+ and WILM-) were calculated by numerical integration of positive and negative powers respectively over time and normalized according to the body mass. The MMA along which moment of the external contact forces is minimal was computed from GRF. was then obtained at each instant of time using: From the second session, the metabolic cost was derived from the oxygen consumption (VO_2 (mL/min)) for each trial. In order to exclusively obtain the metabolic cost for walking the metabolic resting values for standing were subtracted from the metabolic energy consumed during walking. The values were normalized according to the body mass.

IV. RESULTS AND DISCUSSION

A. Comparison of configurations

The average self-selected walking velocity across subject was 1.08 m/s (std: 0.14 km/h). The stiffness of the carbon leaf foot was 203 Nm/rad when participant body mass was inferior to 60kg and 297 Nm/rad when it was superior to 60 kg. The Fig.2, Fig.3 and Fig.4 gives the oxygen consumption, the mechanical work produced by each individual limb (prosthetic and contralateral WILM) during one gait cycle and the evolution of the distance during the gait cycle of the prosthetic limb respectively for the two configurations with the simulator compared to the configuration without prosthesis.

B. Energetic considerations

From the analysis of the oxygen consumption in the three studied configurations, it can be observed that the gait with the simulator induced an increased expense of metabolic energy whatever the ankle-foot used. This trend

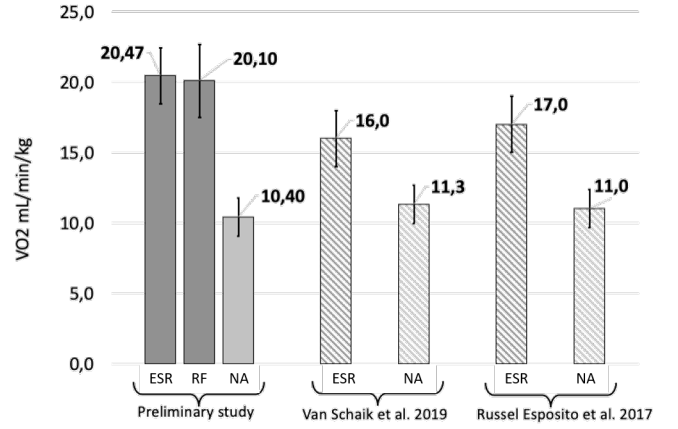


Fig. 2. Average oxygen consumption (VO_2 : mL/min/kg) for the 3 configurations (elastic foot : ESR, rigid foot : RF and non amputee : NA) and comparison to existing literature including transfemoral amputee

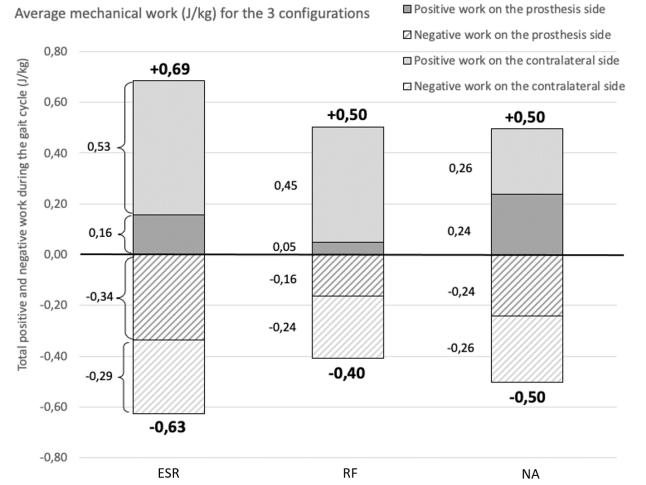


Fig. 3. Average mechanical work WILM (J/kg) for the 3 configurations (elastic foot : ESR, rigid foot : RF and non amputee : NA)

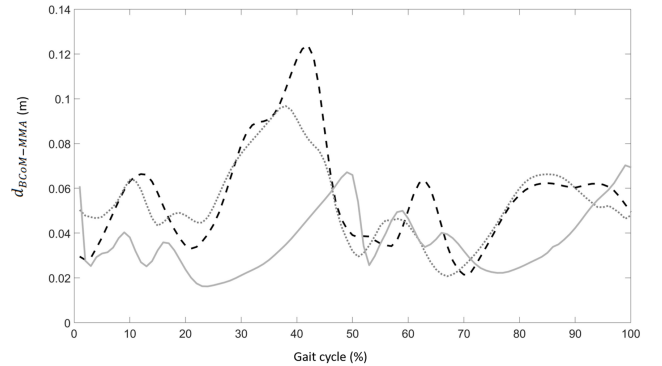


Fig. 4. Average distance between the Body Center of Mass and the Minimal Moment Axis (BCoM-MMA) for the three configurations : without prosthesis in grey solid line, with rigid foot in grey dotted line and with elastic foot in black dashed line

is consistent with the one observed when comparing the energy consumption of people with transfemoral amputation and non-amputee people. However, the oxygen consumption was doubled for all the subjects between the configuration with versus without prosthesis whereas it was reported to be between 30% and 50% between transfemoral amputee and able-bodied persons [38]. In contrast the variation of oxygen consumption between the two prosthetic configurations (elastic and rigid foot) was very limited (2% increase for the elastic foot vs the rigid one). This result can seem counter intuitive considering the property of the elastic foot to store and return energy. Indeed, the elastic foot embeds a carbon leaf that stores mechanical energy by deformation at the beginning of the stance which can be released at the end of the stance when the foot is progressively unloaded during the step-to-step transition. On the contrary, the rigid foot can only roll-over the shape described by the center of pressure in the tibial frame when walking but without any energy storing and return property. The computation of the mechanical work performed by the individual limbs confirmed that positive energy is delivered by the elastic ESR foot (0.16 J/kg) over the gait cycle which is less than the one produced by the ankle of the non-amputee subject (0.25 J/kg) but far more than the one quantified with rigid RF foot (0.05 J/kg). This energy that is provided to the person should theoretically reduce the necessary energy to be produced by the residual joints of the person. However, the analysis of the mechanical work performed by the contralateral limb revealed that an increase of the work performed when wearing the elastic ESR foot (0.53 J/kg) compared to the one performed when wearing the RF foot (0.45 J/kg). Even if the value of the mechanical work cannot be directly extrapolated to explain the increase of the metabolic cost of walking when wearing the simulator, the results of our study are in favor of a role of the contralateral limb in the production of energy in the configuration with the elastic ESR foot. In addition, the analysis of the evolution of over the prosthetic cycle shows an asymmetry of the profile with a significant increase of the distance just before the step-to-step transition at the end of the stance phase. This increase is not retrieved neither at the end of the contralateral stance phase nor for non-amputee people. The values of maximal distance found in our study are consistent with previous studies for both non-amputee persons and people with transfemoral amputation [18]. The increase of the distance in this phase of gait can result from the necessity to unlock the prosthetic knee which requires a muscular action at the residual hip. Even considering the necessity to unlock the knee, it must be also noted that the distance, can be seen as a proxy of the variation of angular momentum, which increase is itself limited by the mechanical actions that can be transmitted at the contact between the foot and the ground (i.e. through friction forces). Finally, it can be hypothesized that it exists a compromise between limiting the instability and unlocking the knee for the swing phase. This compromise is made possible from the muscular actions at different levels of the body that could participate to the increase of the energy consumption and

partly explain the relative inefficiency of the releasing of elastic energy in late stance with the ESR foot.

C. Limitations

Several limitations exist in the present study. The representativity of the simulator has not been evaluated. The generalization of results will necessitate studies including people with amputation. For example, the added mass of the simulator contributes to the increase of the oxygen uptake compared to the “without” configuration. However, it does not explain the overall difference. In the same idea, even if not fully representative of the results that should be obtained with people with amputation, the study shows that energy restitution provided by elastic foot is not sufficient to induce a reduction of metabolic cost of walking. Finally, metabolic cost could not be quantified simultaneously with the biomechanical parameters in the absence of an instrumented treadmill.

V. CONCLUSION

To conclude, the present work is, to our knowledge, the first contribution which aimed to compare elastic energy storing and release foot versus rigid foot from the energetic point of view in femoral prosthetic simulator including a microprocessor-controlled knee. To perform this comparison, three complementary analyses were conducted including metabolic analysis of energy consumption, quantification of the mechanical work performed by the limbs and the estimation of the variation of angular momentum through an indicator developed in the team. The results obtained in the present study show that the mechanical work delivered by the prosthetic foot is not the only determinant of the metabolic energy consumed by the prosthetic user. Considering that the knee is passive and does not provide any energy, this study tends to demonstrate that the overconsumption of energy observed in people with transfemoral amputation also comes from the modification of the dynamics of the overall body. Thus, from our work, it is possible to define targets to be reached by the design and control of the prosthesis and by the rehabilitation process. In particular, the challenge is to manage simultaneously propulsion at the ankle and control of the knee to decrease compensatory motions of the residual limb and the upper body which increase overall instability. This should also result in a decrease of the demand on the contralateral side. The indicators used in the present study are both based on BCoM kinematics and ground reaction forces and moments. One perspective of this work could be to compute them from on board sensors embedded in the device. If the analysis must be deepened to understand the interaction between the foot design and the performed compensation, the results should in the future be also extended to other assistive devices such as exoskeleton and more largely to a better understanding of anthropomorphic action.

ACKNOWLEDGMENT

The authors thank the volunteers who participated in the study.

REFERENCES

- [1] H. (Helen) Huang, J. Si, A. Brandt, and M. Li, "Taking both sides: seeking symbiosis between intelligent prostheses and human motor control during locomotion," *Curr. Opin. Biomed. Eng.*, vol. 20, p. 100314, 2021, doi: 10.1016/j.cobme.2021.100314
- [2] T. W. P. Huang, K. A. Shorter, P. G. Adamczyk, and A. D. Kuo, "Mechanical and energetic consequences of reduced ankle plantarflexion in human walking," *J. Exp. Biol.*, vol. 218, no. 22, pp. 3541–3550, 2015, doi: 10.1242/jeb.113910.
- [3] C. R. Walker, R. R. Ingram, M. G. Hullin, and S. W. McCreath, "Lower limb amputation following injury: a survey of long-term functional outcome," *Injury*, vol. 25, pp. 387–392, 1994, doi: 10.1016/0020-1383(94)90132-5.
- [4] Y. Wen, M. Li, J. Si, and H. Huang, "Wearer-Prosthesis Interaction for Symmetrical Gait: A Study Enabled by Reinforcement Learning Prosthesis Control," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 28, no. 4, pp. 904–913, 2020, doi: 10.1109/TNSRE.2020.2979033.
- [5] H. Houdijk, D. Wezenberg, L. Hak, and A. G. Cutti, "Energy storing and return prosthetic feet improve step length symmetry while preserving margins of stability in persons with transtibial amputation," *J. Neuroeng. Rehabil.*, vol. 15, no. Suppl 1, 2018, doi: 10.1186/s12984-018-0404-9.
- [6] Y. Ding, "Control and Optimization of Soft Exosuit to Improve the Efficiency of Human Walking," Harvard University, 2018. Magn.Jpn., vol. 2, Aug. 1987, pp. 740741 [Dig. 9th Annu. Conf. Magnetics Japan, 1982, p. 301].
- [7] J. M. Donelan, R. Kram, and A. A. D. Kuo, "Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking," *J. Exp. Biol.*, vol. 205, pp. 3717–3727, 2002.
- [8] T. M. Griffin, T. J. Roberts, and R. Kram, "Metabolic cost of generating muscular force in human walking: insights from load-carrying and speed experiments," *J. Appl. Physiol.*, vol. 95, no. 1, pp. 172–183, Jul. 2003, doi: 10.1152/jappphysiol.00944.2002.
- [9] P. a. Willems, G. a. Cavagna, and N. C. Heglund, "External, internal and total work in human locomotion," *J. Exp. Biol.*, vol. 198, pp. 379–393, 1995.
- [10] D. J. Farris and G. S. Sawicki, "The mechanics and energetics of human walking and running: a joint level perspective," *J. R. Soc. Interface*, vol. 9, no. 66, pp. 110–118, 2012, doi: 10.1098/rsif.2011.0182.
- [11] M. M. Arones, M. S. Shourijeh, C. Patten, and B. J. Fregly, "Musculoskeletal Model Personalization Affects Metabolic Cost Estimates for Walking," *Front. Bioeng. Biotechnol.*, vol. 8, no. November, 2020, doi: 10.3389/fbioe.2020.588925.
- [12] J. Donelan, R. Kram, and A. Kuo, "Simultaneous positive and negative external mechanical work in human walking," *J. Biomech.*, vol. 35, pp. 117–124, 2002, doi: 10.1016/S0021-9290(01)00169-5.
- [13] X. Bonnet, C. Villa, I. Loiret, F. Lavaste, and H. Pillet, "Distribution of joint work during walking on slopes among persons with transfemoral amputation," *J. Biomech.*, vol. 129, p. 110843, 2021, doi: <https://doi.org/10.1016/j.jbiomech.2021.110843>.
- [14] B. R. Umberger, "Stance and swing phase costs in human walking," *J. R. Soc. Interface*, vol. 7, no. 50, pp. 1329–1340, 2010, doi: 10.1098/rsif.2010.0084.
- [15] S. M. Bruijn, O. G. Meijer, P. J. Beek, and J. H. Van Dieën, "Assessing the stability of human locomotion: a review of current measures Assessing the stability of human locomotion: a review of current measures," *J. R. Soc. Interface/ R. Soc.*, no. March, 2013, doi: 10.1098/rsif.2012.0999.
- [16] M. Popovic, A. Hofmann, and H. Herr, "Angular momentum regulation during human walking: biomechanics and control," pp. 2405–2411 Vol.3, 2004, doi: 10.1109/robot.2004.1307421.
- [17] F. F. Bailly et al., "A Mechanical Descriptor of Human Locomotion and its Application to Multi-Contact Walking in Humanoids," *Proc. IEEE RAS EMBS Int. Conf. Biomed. Robot. Biomechatronics*, vol. 2018-August, pp. 350–356, Aug. 2018, doi: 10.1109/BIOROB.2018.8488125.
- [18] N. Al-Abiad, H. Pillet, and B. Watier, "A Mechanical Descriptor of Instability in Human Locomotion: Experimental Findings in Control Subjects and People with Transfemoral Amputation," *Appl. Sci.*, vol. 10, no. 840, pp. 1–12, 2020, doi: 10.3390/app10030840.
- [19] J. M. Caputo and S. H. Collins, "Prosthetic ankle push-off work reduces metabolic rate but not collision losses in non-amputee walking experiment," *Sci. Rep.*, vol. 4, p. 7213, 2014.
- [20] P. G. Adamczyk and A. D. Kuo, "Mechanisms of Gait Asymmetry Due to Push-Off Deficiency in Unilateral Amputees," *IEEE Trans. neural Syst. Rehabil. Eng.*, vol. Epub ahead, Sep. 2014, doi: 10.1109/TNSRE.2014.2356722.
- [21] R. C. Riddick and A. D. Kuo, "Mechanical work accounts for most of the energetic cost in human running," *Sci. Rep.*, vol. 12, no. 1, pp. 1–11, 2022, doi: 10.1038/s41598-021-04215-6.
- [22] P. DeVita, J. Helseth, and T. Hortobagyi, "Muscles do more positive than negative work in human locomotion," *J. Exp. Biol.*, vol. 210, no. Pt 19, pp. 3361–3373, 2007, doi: 10.1242/jeb.003970.
- [23] J. R. Montgomery and A. M. Grabowski, "The contributions of ankle, knee and hip joint work to individual leg work change during uphill and downhill walking over a range of speeds," *R. Soc. Open Sci.*, vol. 5, no. 8, 2018, doi: 10.1098/rsos.180550.
- [24] K. E. Zelik and A. D. Kuo, "Mechanical work as an indirect measure of subjective costs influencing human movement," *PLoS One*, vol. 7, no. 2, 2012, doi: 10.1371/journal.pone.0031143.
- [25] H. Houdijk, E. Pollmann, M. Groenewold, H. Wiggerts, and W. Polonski, "The energy cost for the step-to-step transition in amputee walking," *Gait Posture*, vol. 30, no. 1, pp. 35–40, 2009, doi: 10.1016/j.gaitpost.2009.02.009.
- [26] X. Bonnet, C. Villa, P. Fodé, F. Lavaste, and H. Pillet, "Mechanical work performed by individual limbs of transfemoral amputees during step-to-step transitions: Effect of walking velocity," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 228, no. 1, pp. 60–66, 2014, doi: 10.1177/0954411913514036.
- [27] K. E. Zelik and A. D. Kuo, "Human walking isn't all hard work: evidence of soft tissue contributions to energy dissipation and return," *J. Exp. Biol.*, vol. 213, no. 24, pp. 4257–4264, 2010, doi: 10.1242/jeb.044297.
- [28] K. E. Zelik, K. Z. Takahashi, and G. S. Sawicki, "Six degree-of-freedom analysis of hip, knee, ankle and foot provides updated understanding of biomechanical work during human walking," *J. Exp. Biol.*, vol. 218, no. 6, pp. 876–886, 2015, doi: 10.1242/jeb.115451.
- [29] P.-B. Wieber, "On the stability of walking systems," *Proc. Third IARP Int. Work. Humanoid Hum. Friendly Robot.*, pp. 1–7, 2002, [Online].
- [30] P. Sardain and G. Bessonnet, "Forces acting on a biped robot. Center of pressure - Zero moment point," *IEEE Trans. Syst. Man, Cybern. Part A Systems Humans.*, vol. 34, no. 5, pp. 630–637, 2004, doi: 10.1109/TSMCA.2004.832811.
- [31] A. L. Hof, "The 'extrapolated center of mass' concept suggests a simple control of balance in walking," *Hum. Mov. Sci.*, vol. 27, no. 1, pp. 112–125, Feb. 2008, doi: 10.1016/j.humov.2007.08.003.
- [32] A. Goswami, "Postural stability of biped robots and the foot-rotation indicator (FRI) point," *Int. J. Rob. Res.*, vol. 18, no. 6, pp. 523–533, 1999, doi: 10.1177/02783649922066376.
- [33] H. Herr and M. Popovic, "Angular momentum in human walking," *J. Exp. Biol.*, vol. 211, no. 4, pp. 467–481, 2008, doi: 10.1242/jeb.008573.
- [34] R. Neptune and A. Vistamehr, "Dynamic Balance during Human Movement: Measurement and Control Mechanisms," *J. Biomech. Eng.*, vol. 141, no. 7, p. 070801, 2019, doi: 10.1115/1.4042170.
- [35] H. M. Maus, S. W. Lipfert, M. Gross, J. Rummel, and A. Seyfarth, "Upright human gait did not provide a major mechanical challenge for our ancestors," *Nat. Commun.*, vol. 1, no. 6, pp. 1–6, 2010, doi: 10.1038/ncomms1073.
- [36] A. H. Hansen, D. S. Childress, and S. C. Miff, "Roll-over characteristics of human walking on inclined surfaces," *Hum. Mov. Sci.*, vol. 23, no. 6, pp. 807–821, 2004, doi: 10.1016/j.humov.2004.08.023.
- [37] H. Pillet, X. Bonnet, F. Lavaste, and W. Skalli, "Evaluation of force plate-less estimation of the trajectory of the centre of pressure during gait. Comparison of two anthropometric models," *Gait Posture*, vol. 31, no. 2, pp. 147–152, 2010, doi: 10.1016/j.gaitpost.2009.09.014.
- [38] L. van Schaik, J. H. B. Geertzen, P. U. Dijkstra, and R. Dekker, "Metabolic costs of activities of daily living in persons with a lower limb amputation: A systematic review and meta-analysis," *PLoS One*, vol. 14, no. 3, pp. 1–24, 2019, doi: