PASSIVE DYNAMIC SYSTEM FOR ENERGY RETURNING ON TRANSTIBIAL PROSTHESIS

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RESEARCH PROPOSAL

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Summary

Abstract

Nowadays, Lower Limb Prosthesis (LLP) are changing at a very fast pace, due to technological developments implemented in such devices. In addition, users have new demands about their prosthesis and they require absolute comfort and good performance. Unfortunately, the demand of LLP has risen mostly in third world countries because of the increment of vascular diseases (e.g., Diabetes Mellitus). However, people do not have the enough funds to acquire advanced prosthesis that return the capabilities of walking or jogging in a proper way.

Despite the fact that active prosthesis help people to reduce metabolic cost, those are heavier and more expensive than *Energy Storage and Return*(ESR) prosthesis devices, produce uncomfortable noises and require more maintenance than passive ones. Moreover, components of the bionic prosthesis (i.e., actuators, battery, gearbox, among others) make the system highly inefficient. As a consequence, a higher quantity of external energy is required to allow the user having enough autonomy for a daily use.

The current work is a Ph.D. thesis, which purpose is manufacturing a novel customizable configuration of transtibial prosthesis. This device will provide the positive work needed for an amputee at the final stance phase through a passive dynamic system, it will take advantage of cellular solids properties for recycling the energetic lost at the initial contact of the gait.

Nomenclature

AM   Additive Manufacturing

CAD   Computer Aided Design

COM   Center Of Mass

DES   Discrete Element Method

ER   Energetic Requirement

ESR   Energy Storage and Return prosthesis

FEM   Finite Element Method

IC   Initial Contact

IDF   International Diabetes Federation

PP   Peak Power

QoL   Quality of Life

SEA   Series Elastic Actuator

1 MOTIVATION

The demand of Lower Limb Prosthesis (LLP) is higher every day around the world, due to the constant increment of the principal causes of amputations. According to Ziegler *et al.* [1], in the United States of America, amputations in 2008 were caused by vascular diseases (including diabetes) with 53,95%, followed by trauma (e.g. accidents, warfare, among others) with 44,90%, and cancer with 1,15%. They estimated that in 2050 the number of amputees will have risen to 3,6 million [1].

In 2015, the *International Diabetes Federation (IDF)* published the IDF atlas, which announced that the number of people with diabetes is between 340-536 million [2]. Moreover, they estimated that this sickness will have affected 642 million of people worldwide by 2050.

It is believed that diabetes affects mostly the lower limbs, producing potential risks of suffering peripheral arterial illness, diabetic foot and, as a result, an amputation. The possibility of suffering an amputation will depend on race, gender, and age of the population, being different in many countries. Below, Fig. 1 shows the number of amputations per 100.000 habitants caused by diabetes.
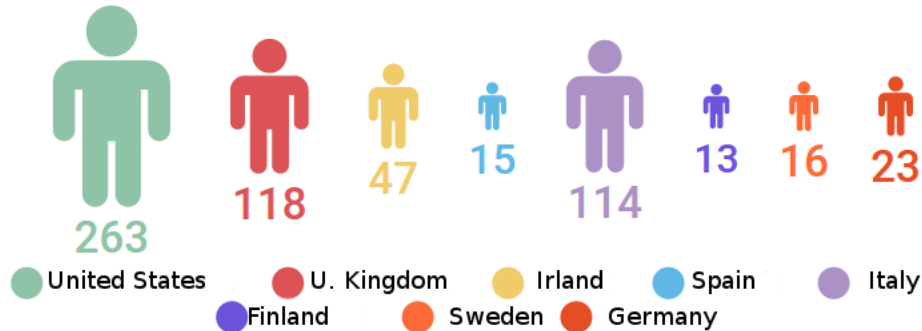

Figure 1: Population affected per 100.000 inhabitants during a specific period of time in each country. Adapted from Kroger and Knut [3].

Despite the lack of statistics of amputations in all countries, diabetes is considered the main cause of amputation around the world.

Even though bionic prosthesis supply more amount of energy than *Energy Storage and Return*(ESR) at final stance phase, those are extremely expensive for developing countries. Thus, we have to take into account those communities with a limited economic capacity to get those devices. In consequence, third world countries sacrifice good Quality of Life (QoL) with low technology prosthesis [4].

To sum up, contributing to the development of prosthetic devices for lower limbs has great relevance for improving disorders in pathological gait and therefore obtain better QoL. The strategy to recover the absence of limbs work is recycling energy of Initial Contact (IC) and return it during final stance phase. It is thought that this strategy could be cheaper than prosthesis with active actuators inside.

2 STATE OF THE ART

2.1 State of the Art about Ankle-Foot Prosthesis

This motivation encourages research on devices that restore energy loss in gait as efficiently as possible. Moreover, replacing ankle joint is a complex task because it should be able to manage its stiffness regardless the terrain or the type of gait. Great technological advances in prosthesis have been made worldwide. There are clearly two kinds of strategies in foot prosthesis: ESR prosthesis, which state-of-the-art was described by Versluys *et al.* [5]; and bionic prosthesis, described by Cherelle *et al.* [6] on what is considered a bionic prosthesis and, consequently, their types. Fig. 2 depicts a generalized representation of foot prosthesis classes, where the most implemented foot prosthesis is the SACH, despite not having returning energy benefits in gait.
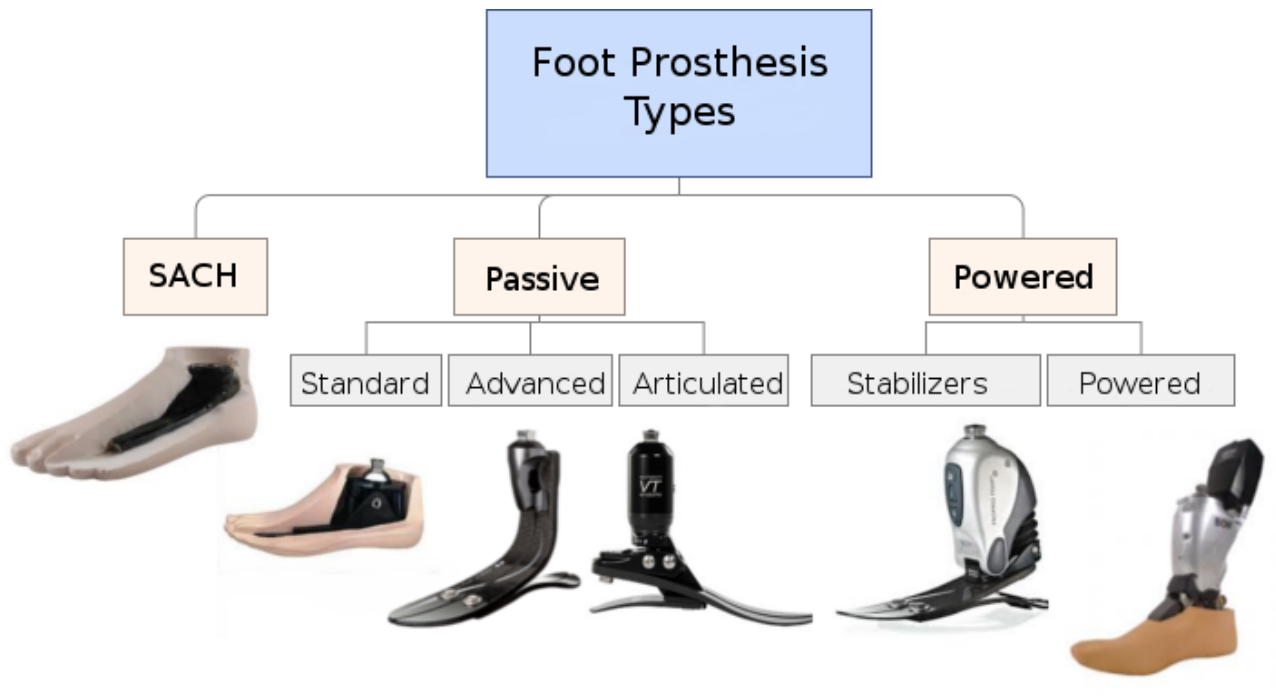

Figure 2: Generalized categorization of ankle-foot prosthesis according to Cherelle *et al.*[6] and Versluys *et al.* [5]. From left to right: SACH foot, sagittal degree of freedom foot, OSSUR® flex foot, Echelon foot®, Proprio foot from OSSUR® and BiOM® from iWalk Inc.

Principal strengths and weaknesses in prosthesis to date have been found, both for ESR and bionic prosthesis, which are depicted in Tables 1, and 3 - 4, respectively.

Furthermore, weaknesses in prosthesis generate biomechanical disorders in gait, which are mentioned in Table 2 for ESR and Table 5 for bionics.

Table 1: Strengths and weaknesses on ESR prosthesis

Passive prosthesis strengths	The price of ESR is cheaper than bionic ones.
	Good source of positive external work at final stance phase without requiring batteries [7].
Passive prosthesis weaknesses	It can only react at final compression of material while actives act and react. [8].
	Significant ankle power difference between the affected and unaffected sides during ankle-powered plantar flexion. [9, 10].
	Poorer shock absorption on affected side [11].
	It cannot replicate the positive work phases of the human joint. [12, 13].
	As prosthetic foot deforms during loading, it will exert a braking effect on the *Center of Mass* (CoM) progression. [14].

Table 2: Biomechanical disorders in ESR prosthesis users

Biomechanical disorders.	Users expend between 20% and 30% more metabolic power to walk. [9, 15].
	Users walk between 30% and 40% slower the same distance in comparison with an able-bodied person. [9, 16, 17, 18, 19, 15].
	Users show assymetric patterns in gait [9, 20, 19].
	Hip extension, knee flexion and ankle dorsiflexion on the unaffected side are higher than normal [21, 22].
	Affected limb has higher stance phase time, higher step length, less swing phase time and less inertia moment than unaffected side.[23].
	People with unilateral transtibial amputation have an increased susceptibility to knee osteoarthritis. [24].
	Biomechanical disorders in gait might be the main cause of dorsal-lumbar pain [25].

In spite of the fact that bionic prosthesis provide more benefits than passive ones in terms of dynamic improvements in gait, nowadays those prosthesis present disadvantages and some dynamic disorders in gait (See Table 5).

Table 3: Bionic prosthesis strengths in comparison to passive ones

Bionic prosthesis strengths	
	The system replaces musculoskeletal work in gait [8].
	It is capable of recognizing different types of terrain and velocities of gait [26].
	It is capable of producing positive mechanical power [12].
	It reduces the metabolic demand up to 16% [17, 13].
	Present more stable trajectory in comparison to passives [19].
	Subjects had a 10% faster self-selected walking speed on uneven ground when wearing the powered compared with ESR. [18].
	Reductions in peak impact resultant force, impact resultant force loading rate, peak heel-strike foot pressure and peak knee external adduction moment when comparing powered ankle-foot prosthesis to the conventional passive prosthesis. [19].
	Decreasing in peak resultant force and adductor knee moments on unaffected limb [24].

Table 4: Recent weaknesses on powered prosthesis

Powered prosthesis weaknesses	
	Electric-based machines make noises. [27].
	Most of them are not scalable, pediatric users cannot use it. [28].
	It still requires biomimetic ankle intervention. [19].
	High cost, up to U$40.000 - U$50.000. Price of BiOM® according to Boston Magazine[27] could be around U$ 40.000. The Proprio® prosthesis from Ossur could be around U$25.000 [29]. On the other hand, an ESR prosthesis is between U$ 500 and U$3.000.
	Most of ESR prosthesis are lighter than powered.
	It requires more maintenance than passive ones.
	Technical parameters need tuning (i.e. power, torque, cycle times, etc) according to anthropometric requirements.
	System is inefficient since mechanical design incurs in energetic losses [27, 6], thus autonomy is lower than in passive prosthesis.
	Specific implementation of control strategies - as Reis mentioned in [30] - are needed for powered prosthesis. Consequently, hardware and software are more demanding.

ESR prosthesis are not able to reestablish the dynamic walking nor the metabolic cost, due to the absence of the lower limb. On the other hand, powered prosthesis have restored the positive work needed to return the controlled energy and satisfy the quasi-stiffness slope [32, 33] of lost ankle. However, methods to reestablish the energy are highly inefficient [6], making those devices more complex and expensive compared to ESR.

Table 5: Recent biomechanical disorders on powered prosthesis

Biomechanical disorders	Since the user gait is more dynamic, the pressure in stump is higher than in ESR users. Hence, the risk of suffering ulcers, dermatitis or any trauma in that area, is higher. [31]
	To date, it only has been designed for adult population. It is not suitable for children [28].

Literature has reported a variety of actuators, which are divided in: stiff actuators and Compliance actuators. The pros and cons are described in Appendix A: State-of-the-art of prosthetic actuators. According to Vanderborght *et al.* [34], those actuators will undergo continuous development aiming to achieve the best energetic efficiency, stability, autonomy and a regular walking.

Even though some actuators have not been made yet, because of the geometric constraints to implement those components, some others have been implemented in commercial prosthesis (e.g. Series Elastic Actuator). That kind of actuators have shown an improvement in walking. Nevertheless, high electric power is required by the actuator, thus autonomy [1] is reduced.

Two variables were taken into account by Grimmer [35, 36] and Eslamy[37, 38] to determine efficiency on actuators. Those are *Peak Power (PP)* and *Energetic Requirement* (ER). The more PP and ER per cycle in the motor, the more electric energy required.

Some actuators mentioned in Appendix A were studied to verify the requirements of the motor, and identify the more efficient. Fig. 3 shows the results of the most popular actuators used in foot prosthesis for a subject of 75 kg of mass and cadence at 1 m/s.

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[1]Autonomy is calculated by number of steps until the powered prosthesis discharges. Generally, It must satisfy between 3000 to 5000 steps per day[24].
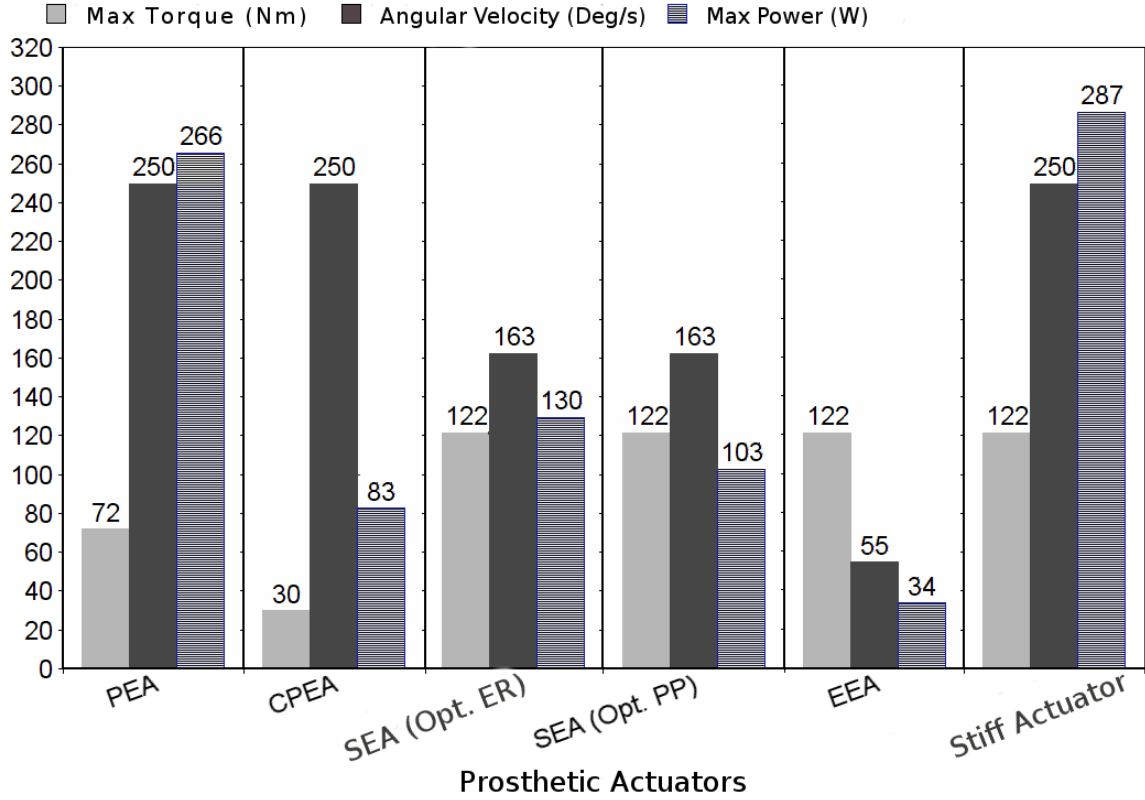
Figure 3: Mechanical requirements of varied prosthetic actuators. Results were obtained by mathematical modeling of each one [6]. Nomenclature: PEA: Parallel Elastic Actuator, CPEA: Clutched PEA, SEA (Opt. ER): SEA actuator focused on optimizing ER, SEA (Opt. PP): SEA actuator focused on optimizing PP, EEA: Explosive Elastic Actuator.

It must be taken into account that the power required for each actuator does not consider energy losses made by the system, which are around 50-60%; Hence, 600W are needed to generate pure power of 287 W by the motor [6].

Based on the foregoing, from the gait analysis, it has been found that energy dissipation occurs when the velocity vector of the COM is redirected at each step-to-step transition [39]. In other words, the stance leg in every gait cycle, acts similarly to an inverted pendulum, in order to support the body COM. Consequently, when transition is made - in double support phase - the COM velocity is redirected and energy is lost at collision of the heel with ground [40]. A graphic explanation can be seen in Fig. 4A.

Therefore, Collins and Kuo [40] built a prosthetic foot which initially recycled the energy lost at initial contact of the gait through the implementation of a compression spring. This spring stores the energy until late stance phase of gait to finally release it, with the purpose of providing positive work to amputees. Nevertheless, neither shock absorption was reduced, nor the system provided

all energy needed to restablish a regular gait.

In general, powered prosthesis follow the next procedure: an ESR foot is used as base, later the actuator is implemented for push-off; Then, a control strategy is needed for its use and finally a battery is installed as source of power. Based on the previous technique for restoring limb loss, a question arised: what other strategy could be implemented in prosthesis so that, it obtains a better efficiency?.

To sum up, pathologies caused by lower limb amputations befall in extra expenditure of energy (i.e. metabolic cost) for patients, due to the absence of *triceps surae* group, which generates approximately 80% of the work needed for push-off. In spite of the fact that ESR prosthesis provide positive work at final stance phase, to date those are not able to provide all the needed work. On the other hand, powered prosthesis are capable of supplying positive work needed by an amputee, but, that type of prosthesis has lower autonomy than ESR since the integrated actuator is inefficient. Hence, the necessity of the design of passive dynamic systems (concept defined by McGeer [41]) for foot prosthesis emerged.

2.2   Additive Manufacturing on Prosthesis

The evolution of technological development in medicine is growing faster nowadays, from manufacturing prototypes for academic purposes to cloning a cell, a tissue or an organ to implant [42]. All the above range of facilities are due to Additive Manufacturing - well known as 3D printing -, which principle is based on making pieces layer by layer, on the basis of CAD models. AM pros and cons in comparison to traditional manufacturing are well described by Weller *et al.*[43] and Diegel *et al.* [44].

Those advantages suited the necessities of prosthesis makers, who demonstrated that mechanical properties of printable materials accomplish the task demanded by a foot prosthesis [45, 46]. It is believed that concepts of prosthesis with AM will adapt new shapes and could possibly change its mechanical properties dynamically, due to the combination of self-assembly technique and 3D printing. Some experiments have been reported by Tibbits *et al.*[47, 48] and Yu *et al.*[49].

Even though few foot prosthesis are made from AM, all of them have uniform mechanical properties all over its structure, This affects negatively the accomplishment of elastic properties needed for satisfying ankle quasi-stiffness on the prosthesis. In addition, given the ability to add material freely in AM without manufacturing penalties, the shape and geometry could be adapted easily in case of changes in the anthropomorphicity of users (i.e. pediatric users).

As a result of the above requirements and facilities given by AM, the use of cellular solids is proposed in order to configure different sub-domains with varied mechanical properties. These materials show some advantages compared to true-solids, such as: i) good thermal conductivity (e.g.
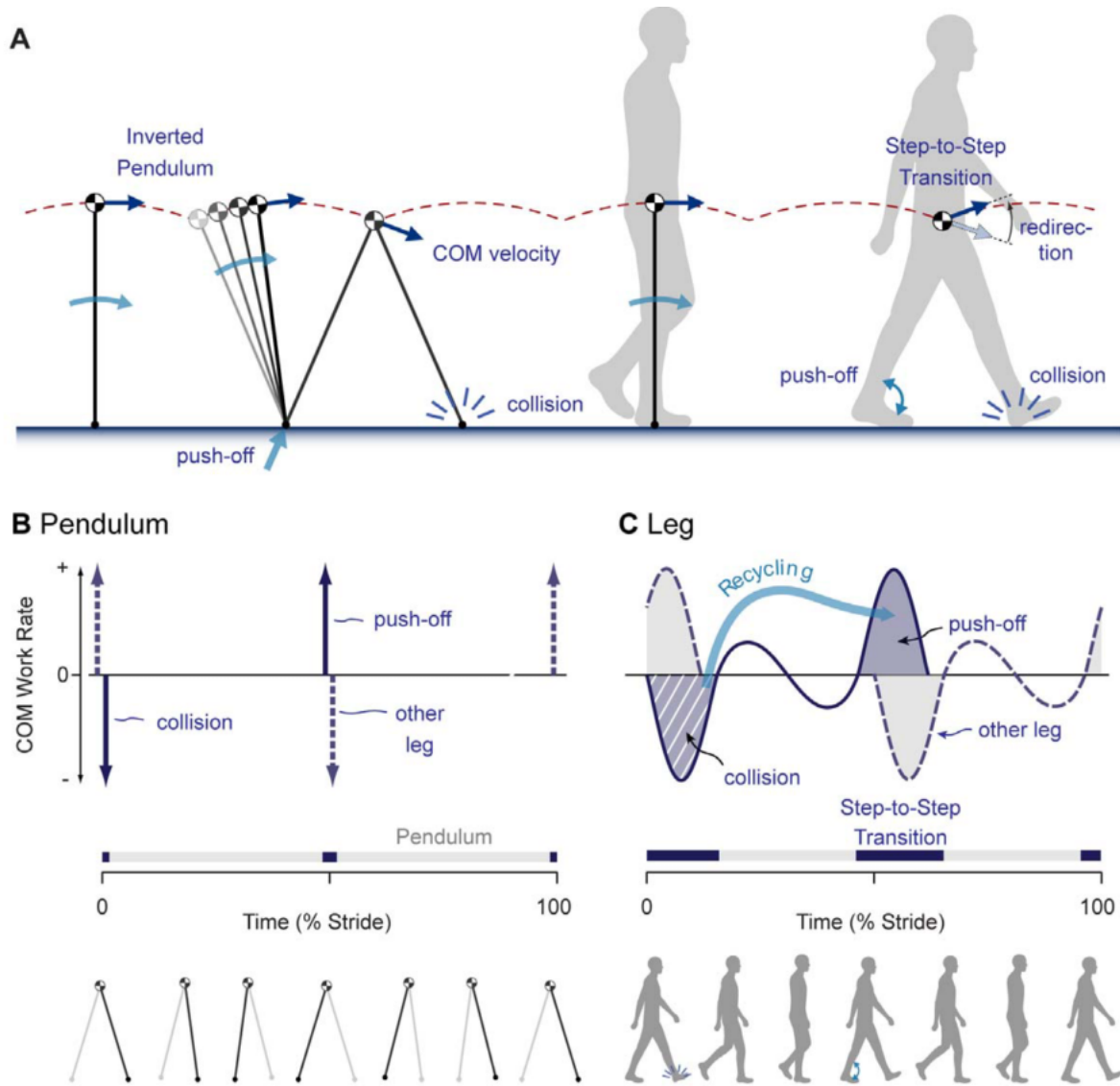
Figure 4: *"(A) The stance leg acts similarly to an inverted pendulum to support the body center of mass. The center of mass velocity is redirected between steps when the other leg contacts the ground with a dissipative collision. (B) The rate of work performed on the center of mass by ideal pendulum-like legs vs. stride time. Work is theoretically minimized by pushing off impulsively (indicated by arrows) just before the opposite leg's collision (step-to-step transition indicated by darkened intervals above time axis). (C) Conceptual plot of center of mass work rate for human-like legs vs. stride time. Imperfectly rigid legs will smooth out the impulses, but the collision (hatched area) is nevertheless a possible source of energy for recycling if it can be captured, stored, and later released for push-off."* Paragraph and figure taken from Collins and Kuo[40].
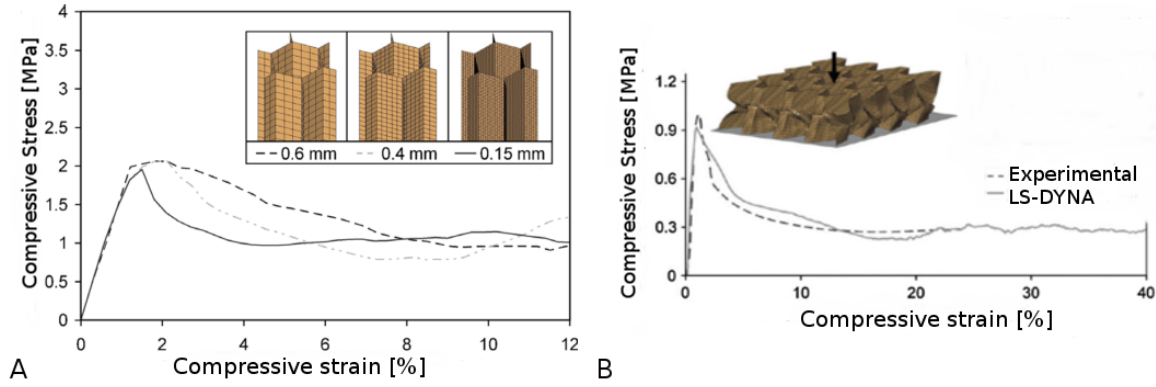
Figure 5: A) Influence of element size (from coarse mesh (0.6 mm) to fine mesh (0.15 mm)) on compressive stress–strain curve for honeycomb model. Reproduced from Heimbs *et al.* B) Validation of virtual testing results vs. experimental testing at compression load. Taken from Heimbs *et al.* [54].

insulation); ii) low weight (e.g. structural sandwich panels); iii) Undergo large deformations (80-90%) at roughly constant (low) stress (e.g. helmets); and iv) Have good mass-strength relationship [50]. Thus, cellular solids can yield a suitable solution of the desired application. For instance, Wang *et al.* [51] configured a meta-material structure that keeps mechanical properties regardless the lengthening of the material and Chu *et al.* [52] investigated design synthesis methods for designing lattice cellular structures to achieve specifically a desired stiffnesses through optimization algorithms.

However, the experimental prediction of the mechanical behaviour of these materials implies a high cost-benefits relationship. Moreover, some structural characteristics such as crystal plasticity, mechanics at different micro-scales, among others, are tediously obtained or cannot be measured by this method [53].

On the other hand, many studies on cellular solids have been made *in silico* through Finite Element Method (FEM), such as Heimbs [54], who made virtual testing of honeycomb sandwich core structures using dynamic FEM simulations varying some parameters on the model (See Figure 5).

3 PROBLEM STATEMENT

- The recent passive prosthesis for transtibial amputees produce disorders in the dynamic parameters of the gait, owing to the absence of positive work of the limb loss.

3.1 Research Question

Which passive ankle-foot prosthesis based on cellular solids configurations, will generate the positive work needed for push-off, taking advantage of the energy lost at initial contact of the gait?

3.2 Hypothesis

A passive dynamic system (compound of cellular solids) within a passive ankle-foot prosthesis, configured to store energy in a controlled manner at initial contact of the early stance phase, will enable to return the stored energy after dorsiflexion phase.

3.3 General and Specific Objectives

3.3.1 General Objective

To suggest an ankle-foot prosthesis being able to generate - through a passive dynamic system - the positive work needed for push-off after dual-flexion phase, taking advantage of the energy lost at initial contact of the gait.

3.3.2 Specific Objecives

1. Identify biomechanical parameters and the work-loop slope of ESR prosthesis users and non-amputees aiming to obtain the ankle quasi-stiffness of both cases.

2. Obtain a preliminary model of the ankle-foot prosthesis capable of storing energy (during initial contact until late dual-flexion phase), and returning it at dorsiflexion phase in a controlled manner through the passive dynamic system.

3. Determine detailed configurations of cellular solids that accomplish the requirements of the preliminary model.

4. Validate the dynamic model of the ankle-foot prosthesis in comparison to an ESR prosthesis.

3.4 Methodology and Activities

The methodological procedure will be carried out through the extraction of statistical data, in order to obtain the ankle quasi-stiffness slope of prosthesis users and non-amputees, so as to compare the energetic gap between those cases. On the other hand, an *in-silico* process to design the prosthesis is needed to propose the concept and verify its functionality. This simulation process will be done with the tentative computational framework proposed in Figure 6.
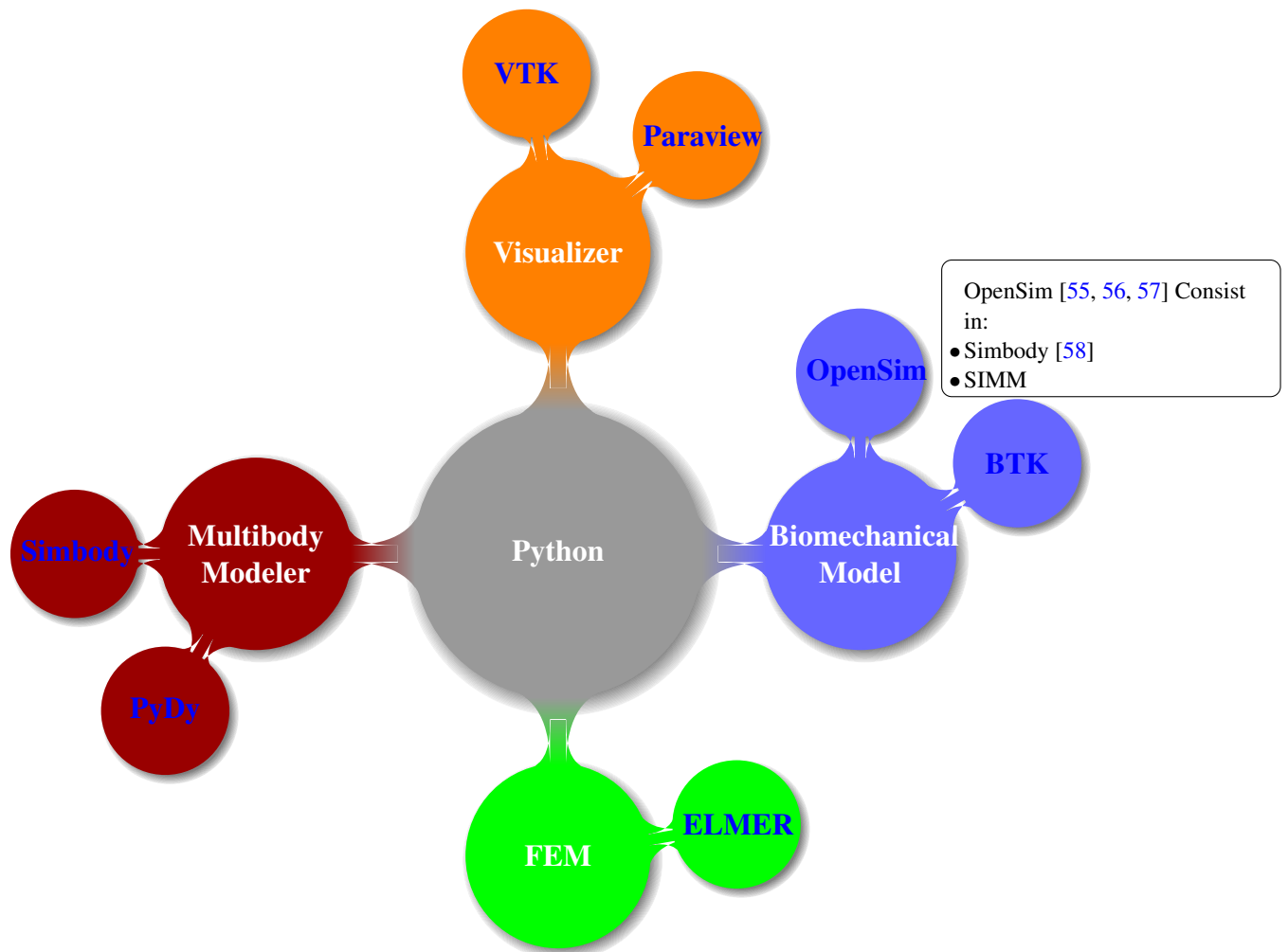
Figure 6: A computational framework of the proposed methodology. Click on the hyperlinks to see more details of every open-source software.

Objective 1: Identify biomechanical parameters and the work-loop slope of ESR prosthesis users and non-amputees aiming to obtain the ankle quasi-stiffness of both cases.

Methodology: Obtaining the most important biomechanical parameters (as mentioned by Sagawa [60]) for ankle quasi-stiffness slope through extraction of data in order to compare the energetic demand of an ESR user with a non-amputee.

Activities:

1. To search in literature the most useful data and filter it.

2. To obtain biomechanical parameters for getting different ankle quasi-stiffness slopes of different ESR and able-bodied patients.
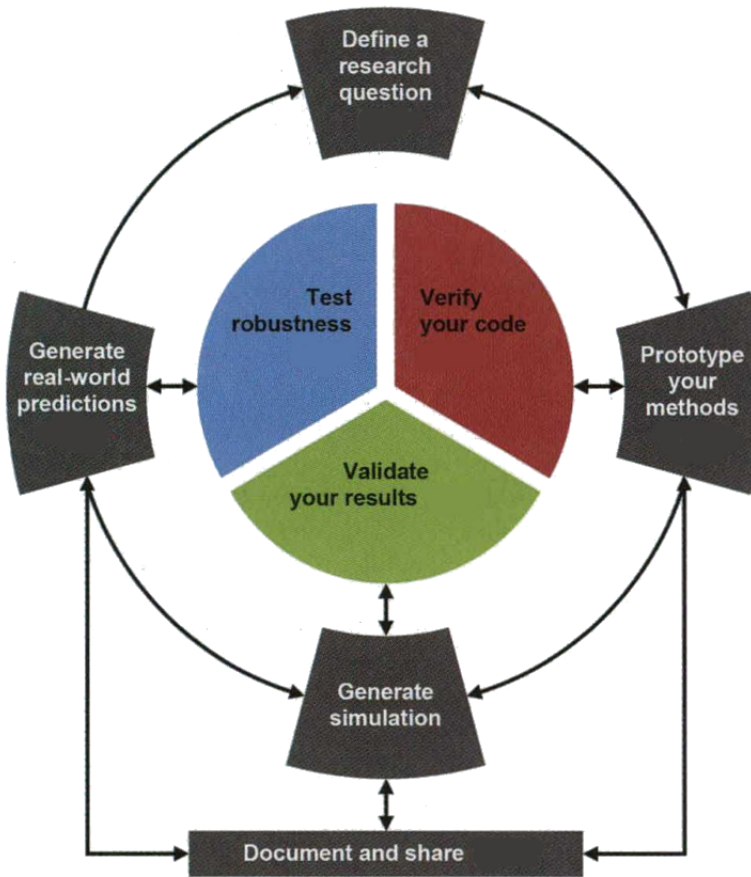
Figure 7: Best validation and verification practices for biomechanical models. Taken from Hicks *et al.* [59]

3. To make the biomechanical model of each subject with the purpose of acquiring non-measurable variables and predicting biomechanical behaviour through forward dynamic technique in specific software like OpenSim®.

4. To obtain ankle quasi-stiffness slope by the combination of kinetic and kinematic variables.

5. To obtain energetic loss in the collision at initial contact of the gait.

6. To verify biomechanical models with similar publications in literature.

Objective 2: Obtain a preliminary model of the ankle-foot prosthesis capable of storing energy (during initial contact until late dual-flexion phase), and returning it at dorsiflexion phase in a controlled manner through the passive dynamic system.

Methodology: To suggest the preliminary model of the ankle-foot prosthesis capable of storing energy, The design of that model will be iterated through FEM tools combined with flexible

multi-body simulations until reaching the optimized configuration of sub-domains conditioned to the anthropometric volume.

Activities:

1. To divide the biomechanical models into gait sub-phases to analyze each one separately.

2. To determine the geometry restricted to the anthropometric volume with CAD software (e.g. Inventor®).

3. To define the elastic configuration needed to store the energy at each sub-phase of gait.

4. To split the global domain into different sub-domains and define constitutive equation of each sub-domain according to the required energetic storage specifications. (PyDy®/Simbody® y ELMER®).

5. To design the mechanism, which stores energy from initial contact to mid-stance phase with the aim of returning it at late stance phase.

6. To verify flexible multi-body models with FEM tools.

Objective 3: To determine detailed configurations of cellular solids that accomplish the requirements of the preliminary model.

Methodology: Departing from the obtained configuration, a cellular material might be applied to each sub-domain to equalize the mechanical requirements of the model obtained in the previous objective.

Activities:

1. To obtain a cellular solid configuration to accomplish the specific elastic properties of each sub-domain.

2. To verify the mechanical behaviour of each sub-domain with its cellular solid configuration.

3. To verify the structural resistance of the entire model submitted to static loads through FEM or DES tools. o FEM.

4. To get the ankle quasi-stiffness slope of the new configuration with the forward dynamics technique in the biomechanical model mentioned above.

Objective 4: Validate the dynamic model of the ankle-foot prosthesis in comparison to an ESR prosthesis.

Methodology: Manufacturing the ankle-foot prosthesis through 3D printing techniques in order to validate the storage and return efficiency. Finally, we will compare with an ESR prosthesis user.

Activities:

1. Accomplish the medical ethic procedure and request permission for the validation process.

2. Select ESR prosthesis users and adapt them to the new prototype.

3. To collect biomechanical parameters of ESR users as well as of the new prototype.

4. To compare the delivered work during the step-to-step transition in gait with the new concept, the standard ESR prosthesis, and able-bodied patients.

5. To determine the energy storage through the strain measurement of each sub-domains by optic equipment like GOM.

6. To evaluate significant biomechanical differences between the new concept of prosthesis and ESR prosthesis on all over the gait.

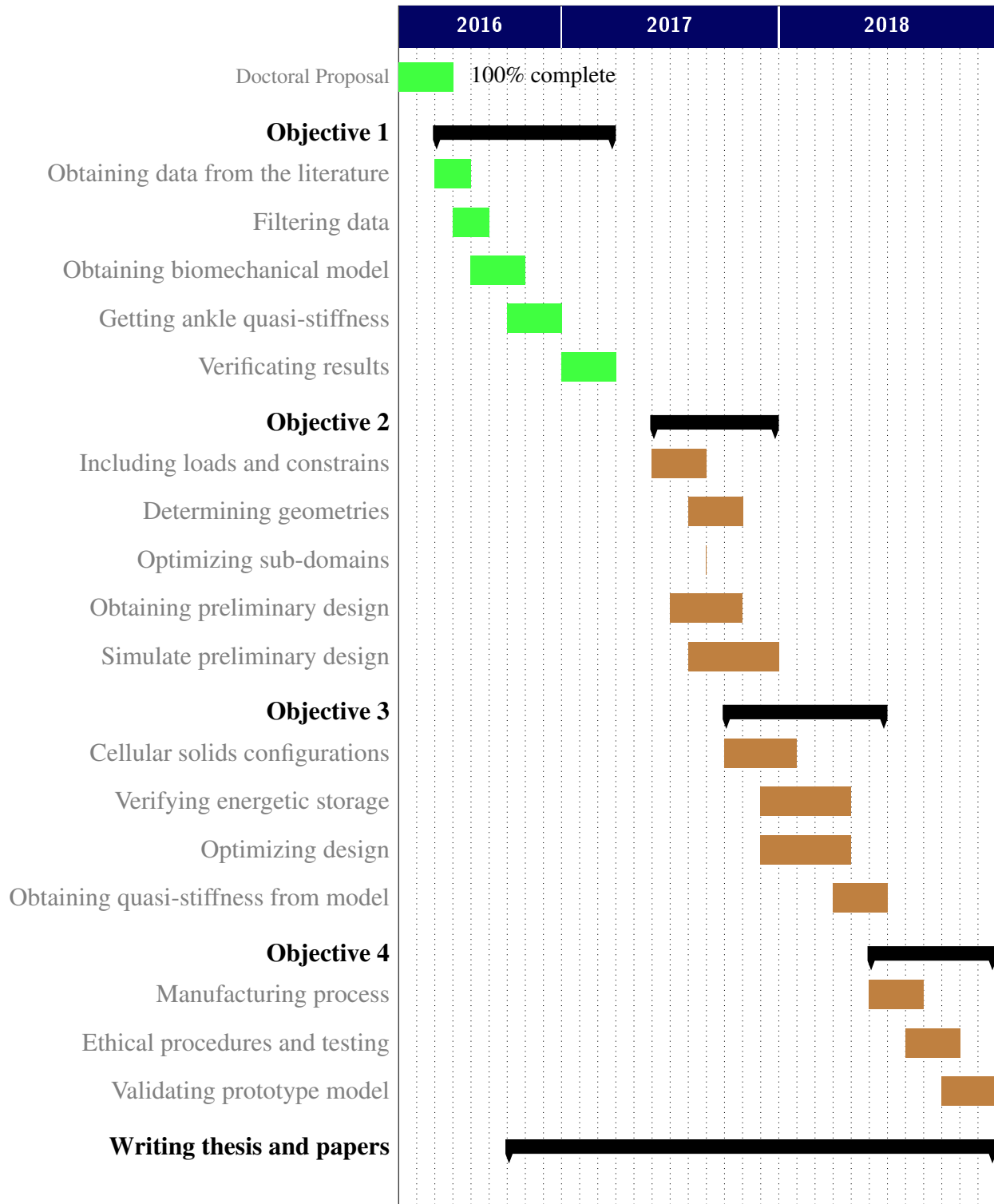7. To conclude results and evaluate future work.

3.5 Implications and Expected Results

The level of impact in this research consist in a new concept of transtibial prosthesis capable of storing energy in many instances of the gait to return it in a controlled manner at final stance phase without the need of external power sources, stepping up energetic contribution in comparison with an ESR prosthesis.

On the other hand, ankle quasi-stiffness depends directly on weight and step length, therefore a customized ankle-foot prosthesis is needed to solve metabolic cost problems in users. In addition, uniform mechanical properties of ESR prosthesis are not able to reproduce the ankle quasi-stiffness properly. The more deficient ankle quasi-stiffness, the more metabolic cost is demanded. In contrast, powered prosthesis generate the positive work needed for an amputee, but its efficiency is much lower than ESR and, as a result, autonomy is reduced. Despite the fact that its electronic components help to get the prosthesis controlled, it increases the cost and third-world countries, where the most amount of amputees live, are not able to acquire one of these.

Moreover, new manufacturing techniques are able to customize the prosthesis according to the anthropometry of users and their specific biomechanical disorders; Hence, we can reach specific elastic properties at different sub-domains through a cellular solid configuration, which have never been seen in any kind of prosthesis.

4 SCHEDULE

	2016	2017	2018
Doctoral Proposal	100% complete		
Objective 1			
Obtaining data from the literature			
Filtering data			
Obtaining biomechanical model			
Getting ankle quasi-stiffness			
Verificating results			
Objective 2			
Including loads and constrains			
Determining geometries			
Optimizing sub-domains			
Obtaining preliminary design			
Simulate preliminary design			
Objective 3			
Cellular solids configurations			
Verifying energetic storage			
Optimizing design			
Obtaining quasi-stiffness from model			
Objective 4			
Manufacturing process			
Ethical procedures and testing			
Validating prototype model			
Writing thesis and papers			

REFERENCES

[1] K. Ziegler-graham, E. J. MacKenzie, P. L. Ephraim, T. G. Travison, R. Brookmeyer, A. Z.-g. K, and M. Ej, "Estimating the Prevalence of Limb Loss in the United States: 2005 to 2050," *Arch Phys Med Rehabil 2008*, vol. 89, no. March, pp. 422–430, 2008.

[2] IDF, "Annual Report 2015," tech. rep., International Diabetes Federation, 2017.

[3] Kroger Knut, "Major and minor amputation rates: What do they tell us?," vol. 15, no. 1, pp. 2014–2016, 2015.

[4] R. LeMoyne, *Advances for Prosthetic Technology*. Tokyo: Springer Japan, 2016.

[5] R. Versluys, P. Beyl, M. Van Damme, A. Desomer, R. Van Ham, and D. Lefeber, "Prosthetic feet: state-of-the-art review and the importance of mimicking human ankle-foot biomechanics.," *Disability and rehabilitation. Assistive technology*, vol. 4, no. 2, pp. 65–75, 2009.

[6] P. Cherelle, G. Mathijssen, Q. Wang, B. Vanderborght, and D. Lefeber, "Advances in Propulsive Bionic Feet and Their Actuation Principles," *Advances in Mechanical Engineering*, vol. 2014, pp. 1–21, 2014.

[7] K. E. Zelik, T.-W. P. Huang, P. G. Adamczyk, and A. D. Kuo, "The role of series ankle elasticity in bipedal walking.," *Journal of theoretical biology*, vol. 346, pp. 75–85, Apr. 2014.

[8] H. A. Varol, F. Sup, and M. Goldfarb, "Multiclass real-time intent recognition of a powered lower limb prosthesis," *IEEE Transactions on Biomedical Engineering*, vol. 57, no. 3, pp. 542–551, 2010.

[9] S. K. Au, J. Weber, and H. Herr, "Powered Ankle Foot Prosthesis Improves Walking Metabolic Economy," vol. 25, no. 1, pp. 51–66, 2009.

[10] H. Herr, J. Weber, K. Au, B. Deffenbaugh, L. Magnusson, A. Hofmann, and B. Aisen, "Powered ankle-foot prosthesis," Mar. 27 2014.

[11] S. Au, M. Berniker, and H. Herr, "Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits," *Neural Networks*, vol. 21, pp. 654–666, May 2008.

[12] E. C. Martinez-Villalpando and H. Herr, "Agonist-antagonist active knee prosthesis: a preliminary study in level-ground walking.," *Journal of rehabilitation research and development*, vol. 46, no. 3, pp. 361–373, 2009.

[13] E. R. Esposito, J. M. a. Whitehead, and J. M. Wilken, "Step-to-step transition work during level and inclined walking using passive and powered ankle-foot prostheses," *Prosthetics and Orthotics International*, 2015.

[14] A. R. De Asha, R. Munjal, J. Kulkarni, and J. G. Buckley, "Impact on the biomechanics of overground gait of using an 'Echelon' hydraulic ankle-foot device in unilateral trans-tibial and trans-femoral amputees.," *Clinical biomechanics (Bristol, Avon)*, vol. 29, pp. 728–34, Aug. 2014.

[15] T. Schmalz, S. Blumentritt, and R. Jarasch, "Energy expenditure and biomechanical characteristics of lower limb amputee gait: The influence of prosthetic alignment and different prosthetic components," *Gait and Posture*, vol. 16, no. 3, pp. 255–263, 2002.

[16] J. G. Buckley, W. D. Spence, and S. E. Solomonidis, "Energy cost of walking: Comparison of 'intelligent prosthesis' with conventional mechanism," *Archives of Physical Medicine and Rehabilitation*, vol. 78, no. 3, pp. 330–333, 1997.

[17] H. M. Herr and A. M. Grabowski, "Powered ankle-foot improves metabolic demand of unilateral transtibial amputees during walking," *Meeting of the American Society of Biomechanics*, 2010.

[18] D. H. Gates, J. M. Aldridge, and J. M. Wilken, "Kinematic comparison of walking on uneven ground using powered and unpowered prostheses.," *Clinical biomechanics (Bristol, Avon)*, vol. 28, pp. 467–72, Apr. 2013.

[19] D. Hill and H. Herr, "Effects of a powered ankle-foot prosthesis on kinetic loading of the contralateral limb: A case series," *IEEE International Conference on Rehabilitation Robotics*, 2013.

[20] E. C. Martinez-Villalpando, L. Mooney, G. Elliott, and H. Herr, "Antagonistic active knee prosthesis. A metabolic cost of walking comparison with a variable-damping prosthetic knee," *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 8519–8522, 2011.

[21] D. C. Morgenroth, A. D. Segal, K. E. Zelik, J. M. Czerniecki, G. K. Klute, P. G. Adamczyk, M. S. Orendurff, M. E. Hahn, S. H. Collins, and A. D. Kuo, "The effect of prosthetic foot push-off on mechanical loading associated with knee osteoarthritis in lower extremity amputees," *Gait & Posture*, vol. 34, pp. 502–507, oct 2011.

[22] H. Bateni and S. J. Olney, "Kinematic and Kinetic Variations of Below-Knee Amputee Gait," *JPO Journal of Prosthetics and Orthotics*, vol. 14, no. 1, pp. 2–10, 2002.

[23] S. J. Mattes, P. E. Martin, and T. D. Royer, "Walking symmetry and energy cost in persons with unilateral transtibial amputations: Matching prosthetic and intact limb inertial properties," *Archives of Physical Medicine and Rehabilitation*, vol. 81, pp. 561–568, May 2000.

[24] A. M. Grabowski and S. D. Andrea, "Effects of a powered ankle-foot prosthesis on kinetic loading of the unaffected leg during level-ground walking," *Journal of NeuroEngineering and Rehabilitation*, 2013.

[25] H. Devan, P. Hendrick, D. C. Ribeiro, L. A Hale, and A. Carman, "Asymmetrical movements of the lumbopelvic region: Is this a potential mechanism for low back pain in people with lower limb amputation?," *Medical Hypotheses*, vol. 82, no. 1, pp. 77–85, 2014.

[26] B. E. Lawson, H. A. Varol, and M. Goldfarb, "Ground adaptive standing controller for a powered transfemoral prosthesis," *IEEE International Conference on Rehabilitation Robotics*, 2011.

[27] BostonNews, "A Brand-New Kick: The New BiOM Ankle Prosthetic by MIT's Hugh Herr," March 2015.

[28] ASME, "Making Strides with Bionic Ankle," March 2015.

[29] R. Jena, "Ossur: Design that walks the line," July 2007.

[30] L. P. Reis and C. P. Santos, "Robot 2015: Second Iberian Robotics Conference," vol. 418, pp. 209–220, 2016.

[31] S. I. Wolf, M. Alimusaj, L. Fradet, J. Siegel, and F. Braatz, "Pressure characteristics at the stump/socket interface in transtibial amputees using an adaptive prosthetic foot.," *Clinical biomechanics (Bristol, Avon)*, vol. 24, pp. 860–5, Dec. 2009.

[32] A. Hansen and E. Nickel, "Ankle-foot prosthesis for automatic adaptation to sloped walking surfaces," Mar. 27 2014.

[33] A. Eshraghi, N. A. A. Osman, H. Gholizadeh, S. Ali, and B. Shadgan, "100 Top-Cited Scientific Papers in Limb Prosthetics.," *Biomedical engineering online*, vol. 12, no. 1, p. 119, 2013.

[34] B. Vanderborght, A. Albu-Schaeffer, A. Bicchi, E. Burdet, D. Caldwell, R. Carloni, M. Catalano, O. Eiberger, W. Friedl, G. Ganesh, M. Garabini, M. Grebenstein, G. Grioli, S. Haddadin, H. Hoppner, A. Jafari, M. Laffranchi, D. Lefeber, F. Petit, S. Stramigioli, N. Tsagarakis, M. Van Damme, R. Van Ham, L. Visser, and S. Wolf, "Variable impedance actuators: A review," *Robotics and Autonomous Systems*, vol. 61, pp. 1601–1614, Dec. 2013.
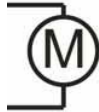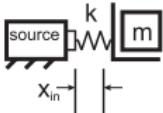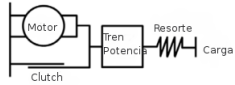
[35] M. Grimmer, M. Eslamy, S. Gliech, and A. Seyfarth, "A comparison of parallel- and series elastic elements in an actuator for mimicking human ankle joint in walking and running," *Proceedings - IEEE International Conference on Robotics and Automation*, pp. 2463–2470, 2012.

[36] M. Grimmer, M. Eslamy, and A. Seyfarth, "Energetic and Peak Power Advantages of Series Elastic Actuators in an Actuated Prosthetic Leg for Walking and Running," *Actuators*, vol. 3, no. 1, pp. 1–19, 2014.

[37] M. Eslamy, M. Grimmer, S. Rinderknecht, and A. Seyfarth, "Does it pay to have a damper in a powered ankle prosthesis? A power-energy perspective.," *IEEE ... International Conference on Rehabilitation Robotics : [proceedings]*, vol. 2013, p. 6650362, June 2013.

[38] M. Eslamy, M. Grimmer, and A. Seyfarth, "Adding passive biarticular spring to active mono-articular foot prosthesis: Effects on power and energy requirement," in *2014 IEEE-RAS International Conference on Humanoid Robots*, pp. 677–684, Nov 2014.

[39] J. M. Donelan, R. Kram, and A. D. Kuo, "Simultaneous positive and negative external mechanical work in human walking," *Journal of Biomechanics*, vol. 35, no. 1, pp. 117–124, 2002.

[40] S. H. Collins and A. D. Kuo, "Recycling energy to restore impaired ankle function during human walking," *PLoS ONE*, vol. 5, no. 2, 2010.

[41] T. McGeer, "Passive dynamic walking," *The international journal of robotics research*, vol. 9, no. 2, pp. 62–82, 1990.

[42] M. H. Michalski and J. S. Ross, "The Shape of Things to Come," *JAMA*, vol. 312, p. 2213, dec 2014.

[43] C. Weller, R. Kleer, and F. T. Piller, "Economic Implications of 3D printing: Market structure Models in light of additive manufacturing Revisited," *International Journal of Production Economics*, Mar. 2015.

[44] O. Diegel, *Additive Manufacturing: An Overview*, vol. 10. Elsevier, 2014.

[45] B. J. South, N. P. Fey, G. Bosker, and R. R. Neptune, "Manufacture of energy storage and return prosthetic feet using selective laser sintering.," *Journal of biomechanical engineering*, vol. 132, no. 1, p. 015001, 2010.

[46] J. Yap and G. Renda, "Low-cost 3D-printable Prosthetic Foot," in *European Conference Design4Health*, (Sheffield), pp. 1–10, Design4Health, 2015.

[47] S. Tibbits, "4D printing: Multi-material shape change," *Architectural Design*, vol. 84, pp. 116–121, 2014.

[48] D. Raviv, W. Zhao, C. McKnelly, A. Papadopoulou, A. Kadambi, B. Shi, S. Hirsch, D. Dikovsky, M. Zyracki, C. Olguin, R. Raskar, and S. Tibbits, "Active Printed Materials for Complex Self-Evolving Deformations," *Scientific Reports*, vol. 4, p. 7422, 2014.

[49] K. Yu, A. Ritchie, Y. Mao, M. L. Dunn, and H. J. Qi, "Controlled Sequential Shape Changing Components by 3D Printing of Shape Memory Polymer Multimaterials," *Procedia IUTAM*, vol. 12, pp. 193–203, 2015.

[50] L. J. Gibson and M. F. Ashby, *Cellular solids. Structure and properties*. Cambridge University Press, 1997.

[51] K. Wang, Y. Zhao, Y. H. Chang, Z. Qian, C. Zhang, B. Wang, M. A. Vannan, and M. J. Wang, "Controlling the mechanical behavior of dual-material 3D printed meta-materials for patient-specific tissue-mimicking phantoms," *Materials and Design*, vol. 90, pp. 704–712, 2016.

[52] J. Chu, S. Engelbrecht, G. Graf, and D. W. Rosen, "A Comparison of Synthesis Methods for Cellular Structures with Application to Additive Manufacturing," *Rapid Prototyping Journal*, vol. 16, no. 4, pp. 459–472, 2010.

[53] M. I. Okereke, A. I. Akpoyomare, and M. S. Bingley, "Virtual testing of advanced composites, cellular materials and biomaterials: A review," *Composites Part B: Engineering*, vol. 60, pp. 637–662, 2014.

[54] S. Heimbs, "Virtual testing of sandwich core structures using dynamic finite element simulations," *Computational Materials Science*, vol. 45, pp. 205–216, apr 2009.

[55] S. L. Delp, F. C. Anderson, A. S. Arnold, P. Loan, A. Habib, C. T. John, E. Guendelman, and D. G. Thelen, "OpenSim: Open-source software to create and analyze dynamic simulations of movement," *IEEE Transactions on Biomedical Engineering*, vol. 54, no. 11, pp. 1940–1950, 2007.

[56] J. a. Reinbolt, A. Seth, and S. L. Delp, "Simulation of human movement: Applications using OpenSim," *Procedia IUTAM*, vol. 2, pp. 186–198, 2011.

[57] A. Seth, M. Sherman, J. a. Reinbolt, and S. L. Delp, "OpenSim: A musculoskeletal modeling and simulation framework for in silico investigations and exchange," *Procedia IUTAM*, vol. 2, pp. 212–232, 2011.

[58] M. A. Sherman, A. Seth, and S. L. Delp, "Simbody: Multibody dynamics for biomedical research," *Procedia IUTAM*, vol. 2, pp. 241–261, 2011.

[59] J. L. Hicks, T. K. Uchida, A. Seth, A. Rajagopal, and S. Delp, "Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of human movement.," *Journal of biomechanical engineering*, vol. 137, no. 2, p. 020905, 2014.

[60] Y. Sagawa, K. Turcot, S. Armand, A. Thevenon, N. Vuillerme, and E. Watelain, "Biomechanics and physiological parameters during gait in lower-limb amputees: a systematic review.," *Gait & posture*, vol. 33, pp. 511–26, Apr. 2011.

[61] Ossur Corp., "PROPRIO FOOT OSSUR," 2014.

[62] D. Paluska and H. Herr, "Series elasticity and actuator power output," *Proceedings 2006 IEEE International Conference on Robotics and Automation, 2006. ICRA 2006.*, no. May, pp. 1830–1833, 2006.

[63] M. a. Holgate, J. K. Hitt, R. D. Bellman, T. G. Sugar, and K. W. Hollander, "The SPARKy (spring ankle with regenerative kinetics) project: Choosing a DC motor based actuation method," *Proceedings of the 2nd Biennial IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics, BioRob 2008*, pp. 163–168, 2008.

[64] R. D. Bellman, M. a. Holgate, and T. G. Sugar, "SPARKy 3: Design of an active robotic ankle prosthesis with two actuated degrees of freedom using regenerative kinetics," *Proceedings of the 2nd Biennial IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics, BioRob 2008*, pp. 511–516, 2008.

[65] F. Sup, H. A. Varol, J. Mitchell, T. Withrow, and M. Goldfarb, "Design and Control of an Active Electrical Knee and Ankle Prosthesis.," *Proceedings of the ... IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics. IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics*, vol. 2008, pp. 523–528, Oct. 2008.

[66] F. Sup, H. A. Varol, J. Mitchell, T. J. Withrow, and M. Goldfarb, "Preliminary evaluations of a self-contained anthropomorphic transfemoral prosthesis," *IEEE/ASME Transactions on Mechatronics*, vol. 14, no. 6, pp. 667–676, 2009.

[67] K. H. Ha, H. A. Varol, and M. Goldfarb, "Volitional control of a prosthetic knee using surface electromyography.," *IEEE transactions on bio-medical engineering*, vol. 58, pp. 144–51, Jan. 2011.
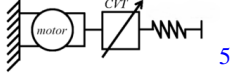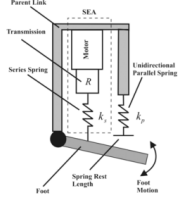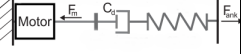
[68] E. J. Rouse, L. M. Mooney, and H. M. Herr, "Clutchable series-elastic actuator: Implications for prosthetic knee design," *The International Journal of Robotics Research*, vol. 33, no. 13, pp. 1611–1625, 2014.

[69] E. J. Rouse, L. M. Mooney, E. C. Martinez-Villalpando, and H. M. Herr, "Clutchable series-elastic actuator: Design of a robotic knee prosthesis for minimum energy consumption," *IEEE International Conference on Rehabilitation Robotics*, no. 1122374, 2013.

[70] L. Mooney and H. Herr, "Continuously-variable series-elastic actuator.," *IEEE ... International Conference on Rehabilitation Robotics : [proceedings]*, vol. 2013, p. 6650402, June 2013.

[71] H. Herr, R. Casler, Z. Han, C. Barnhart, G. Girzon, and D. Garlow, "Controlling torque in a prosthesis or orthosis based on a deflection of series elastic element," Dec. 1 2011.

[72] Z. Han, C. Barnhart, D. Garlow, A. Bolger, H. Herr, G. Girzon, R. Casler, and J. McCarthy, "Controlling powered human augmentation devices," Oct. 11 2012.

[73] Z. Han, C. Barnhart, D. Garlow, A. Bolger, H. Herr, G. Girzon, J. McCarthy, and R. Casler, "Controlling powered human augmentation devices," Mar. 27 2014.

[74] T. Wahl and K. Berns, *Modeling, Simulation and Optimization of Bipedal Walking*, vol. 18. 2013.

[75] P. Cherelle, K. Junius, V. Grosu, H. Cuypers, B. Vanderborght, and D. Lefeber, "The AMP-Foot 2.1 : actuator design, control and experiments with an amputee.," *Robotica*, no. September 2014, pp. 1–15, 2014.

[76] P. Cherelle, V. Grosu, A. Matthys, B. Vanderborght, and D. Lefeber, "Design and validation of the ankle mimicking prosthetic (AMP) Foot 2.0," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 22, no. 1, pp. 138–148, 2014.

[77] J. Geeroms, L. Flynn, R. Jimenez-Fabian, B. Vanderborght, and D. Lefeber, "Ankle-Knee prosthesis with powered ankle and energy transfer for CYBERLEGs α-prototype," *IEEE International Conference on Rehabilitation Robotics*, 2013.

Appendix A: State-of-the-art of prosthetic actuators

Type of Actuator	Graphic Representation	Advantages	Disadvantages	Implemented in
Direct Drive (Stiff Actuator)		It generates positive power at swing phase in order to produce stability [61].	This device is not able to produce positive work at final stance phase[10].	Proprio Ossur ®
SEA[2]		*"The series elasticity changes the operating velocity of the actuator and therefore changes the work and power output"*[62].	*"Issues may include actuator output mass, nonlinear(hardening) series elasticity, nonlinear force-velocity limitations, force-displacement limitations and actuator efficiency."* Additionally, muscles have higher efficiency at low velocities, meanwhile electric motors have higher efficiency at high velocities.[62].	-SPARKy (*Spring Ankle with Regenerative Kinetics*)[63, 64]. -Transfemoral prosthesis prototype by Vanderbilt University [65, 66, 8, 67].
CSEA[3]		During stance phase, the clutch is employed to store elastic energy inside the actuator[68], in turn, a reduced energy (about 70%) was needed to propel the mass compared to conventional SEA [69].	Besides the same disadvantages of SEA, invariable stiffness does not allow providing maximum impedance to actuators. [68].	iWalk knee prototype®.

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[2]SEA: Series Elastic Actuator.
[3]CSEA: Clutchable Series Elastic Actuator

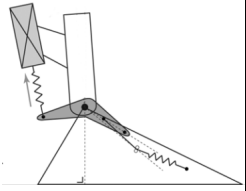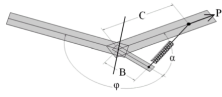| Type of Actuator	Graphic Representation	Advantages	Disadvantages	Implemented in
CV-SEA[4]		It includes a gear box inside actuator so that it can reduce torque sent by DC motor, thus the actuator might increase its energetic efficiency. Moreover, CV-SEA optimizes the required velocity profile during gait, decreasing energy loss.[70].	It has not been implemented in lower limb prosthesis.	
SEAPS[6]		This actuator has the advantage of changing torque and velocity independently at the same time [6].	The same mentioned above.	BiOM®. prosthesis by iWalk®.[71, 10, 72, 73]
SEDA[7]		This actuator requires the less PP at down stair walking. [37].	It does not have any other benefit in walking gait in comparison with other simpler actuators.[37]	Not implemented to date.
PEDA[8]		It was found that this actuator requires less motor force compared to SEA at the beginning of stance phase.[37]	Energetic requirements are increased at the same torque.[37]	Not implemented to date.

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[4]CVSEA: Continuously Variable Series Elastic Actuator

[5]CVT: Continuously Variable Transmission.

[6]SEAPS: Series Elastic Actuator with Parallel Spring

[7]SEDA:Series Elastic Damper Actuator.

[8]PEDA: Parallel Elastic Damper Actuator.

Type of Actuator	Graphic Representation	Advantages	Disadvantages	Implemented in
EEA[9]		Principle of optimal distribution power is implemented in this actuator[74]. Produce up to 3.3 W/kg of PP with an energetic motor requirement of 60 W. Acts during all stance phase [75].	There are some kinematic variations at ankle joint. It is not anthropometric [76].	AMP-foot 2.0 [76] y 2.1 [75].
VSA[10]		It is a robotic system able to change joint stiffness through the thightening of springs. Capable to provide up to 100% of the propulsion power needed.[6]	It requires two motors for its operation, hence more electric power is required[77]. The system is heavier and more complex in technological terms than SEA[6].	CYBERLEG Project (Prototype)