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SURVEY PAPER

Literature review and current trends on transfemoral powered prosthetics

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ABSTRACT
Transfemoral amputation is a common amputation procedure for the human lower limbs. Passive, semi-active, and active prosthetic devices are usually prescribed to amputees in order to restore their quality of life (QOL) according to their abilities. From an engineering perspective, prosthetic and normal gait analysis, actuator technology, and biologic prosthetic control strategies are some of the current objects of study on active lower limb prosthetic design which aimed to reduce potential biomechanical disorders such as gait asymmetry or elevated metabolic cost due to the use of passive prosthetic devices. The main goal of active prosthetic design is to deliver prosthetic assistance at biological levels. This paper reviews the latest developments of semi-active and active prosthetics for transfemoral amputees; as well as the common design considerations and efficiency assessments performed on active transfemoral prosthetics under development in recent years.

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1. Introduction

Many external prosthetic devices have been part of the history since ancient times with aesthetical or functional purposes. The oldest evidence of prosthetics dates back to the ancient Egypt, with the Greville Chester and the Tabeketenmut toes, dating back to 600 B.C. and a period between 950 and 710 B.C., respectively. Other prosthetics for upper and lower limbs made of different materials and mechanisms have been part of history subsequently.

Transfemoral prosthetics are prescribed for individuals who have been amputated in the long section of the femur due to trauma or vascular diseases. These types of devices include knee and ankle artificial joints and are classified according to their actuator schemes as passive, semi-active, and active devices.

Passive prosthetics are operated through mechanical interactions between the residual limb and the terrain along the prosthetic socket and structure. The joint mechanisms on passive prosthetic devices could vary from single axis to polycentric joints on the knee using manual or weight-activated stance control. The swing phase could be controlled using constant or variable friction mechanisms or with pneumatic/hydraulic actuators. Whereas foot prosthetics could be prescribed as SACH (Solid-Ankle Cushion-Heel), simple/multiple axis or dynamic response feet. The types of mechanisms on each transfemoral prosthetic is prescribed by a prosthetist expert according to the strength and ability of the amputee. Passive prosthetics are designed mainly to provide stability during the stance phase of the gait cycle.

Semi-active prosthetics for lower limbs are mainly controlled by microprocessors to assist the amputee during both the swing and the stance phase of the gait cycle. These prostheses are controlled by means of a variety of sensorial data such as inertial, force, angular or electromyographic. These common signals are used on the control systems to improve stability while modulating joint stiffness for different gait speed and terrain. However, these types of prostheses are limited to modulate joint stiffness during the stance phase of the gait cycle in order to improve walking stability and to transfer positive mechanical work to the joints. The development of semi-active lower limb prosthetics began in the early 1990s allowing great advances on gait analysis.

In this context, the term 'active' refers to a prosthesis, whose principal actuation mechanism for the joints is externally powered (i.e. by an electric motor) and controlled through embedded sensors to assist the amputee on activities of daily living (ADL). In general, the prosthesis supplies positive work on at least one of the joints. Most of the progress on active lower limb prosthetics has been developed since the early 2000's up to date as a result of the developments on robotic actuators and control systems.

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From a mechanics standpoint, the human-gait is a combination of muscle activity and synergic movements which makes it a complex activity. Passive transfemoral prosthetics have limited capacity in restoring the functionality of a lost extremity. A proof of this statement is the increased levels of metabolic cost reported by literature [1–5], where several tests have proven that transfemoral amputees require up to 60% more oxygen to perform level walking compared to non-amputees. In some cases, lower limbs prosthetics can cause gait asymmetries which could lead to musculoskeletal disorders both in the residual or the contralateral limb due to the absence of external power for ADL that involve the elevation of the body weight (ascending slopes or stairs).

This study provides an overview on the current state of the art on active transfemoral prosthetics. The topics covered in this paper include the main features studied in recent years regarding design and analyses of prosthetic gait, actuation, and control systems. In addition, the background of modern transfemoral semi-active and active prostheses under development nowadays and the description of performance assessments made in some active prototypes is presented. Finally, a summary of the current trends on active prosthetic design for lower limb amputees is discussed.

2. Design of active lower limb prosthetics

The development of 'smart' (semi-active and active) lower limb prosthetics is mainly based on three principal areas in order to improve stability during prosthetic gait: gait analysis, mechanical design, and control systems. Table 1 describes some of the common considerations regarding on these three areas of lower limb prosthetics development. The approaches and issues addressed by different authors for these three main areas are briefly described in this section regarding the development of active transfemoral prosthetics in recent years.

2.1. Experimental data: gait analysis

Tests on gait analysis have allowed improvements in the development of active prosthetic devices through the gait cycle characterization, mainly for slope and level walking, while stair ascent/descent have been studied in a least extent. Quantitative gait analysis comprise gait data of kinematics, kinetics, electromyography (EMG), and metabolic cost [6].

Most of the amputee gait studies available in the literature have been performed on similar tests for level surfaces and ramps at different slopes [7], while gait analyses for transfemoral amputees during stair ascent/descent have shown differences between diverse studies on their experimental setups. These inconsistencies could be due to the difficulties for lower limb amputees (particularly transfemoral) to ascend stairs on a step-over-step manner as a consequence of the lack of positive mechanical work, which is required to compensate the body weight when the prosthetic limb is on the stance phase. For example, Bae et al. [8] requested amputee volunteers to perform stair ascent on a step-by-step manner if they were not able to ascend step-over-step on a daily basis; other authors have performed tests on non-amputee subjects during stair gait [9–11], while Hobara et al. [12] asked volunteers to ascend step-over-step even if they were not able to do it on a daily basis.

In contrast, other experimental conditions have been consistent among researchers for stair gait, such as muscle selection for surface EMG detection [8,10,13–15] and the number of steps for stair gait analysis. For the latter, authors have concurred in a minimum of five steps on their stairs with at least two steps sensing ground reaction forces to gather data of at least a full gait cycle in steady state [16–20].

Peng et al. studied gait transitions of non-amputee volunteers with kinematic and EMG data during level walking to stair ascent and descent [21]. The experiments were recorded up to three strides before each transition to stair gait. The main features observed between the leading and the trailing leg were some changes in the velocity (on the last stride before transition to stairs) and the myoelectric activity of plantar-flexor muscles.

2.1.1. Kinematics

Kinematic pattern recordings on lower limb joints during the gait cycle under laboratory conditions have been

Table 1. Common considerations on the development of semi-active and active lower limb prosthetics.

Development of smart prosthetics	Experimental data	Type of gait	Level, slope, stairs, running, jumping, cycling, etc
		Gait analysis data	Kinematics, kinetics, metabolic cost, electromyography
	Biomechanical design	Structural design	Geometry, materials, assembly, strength tests
		Biomechanical compliance	Range of motion, velocity, torque and power requirements of artificial knee and ankle joints
		Actuator technology	Electro-mechanical, hydraulics, pneumatics, magneto/electro rheology
	Control systems	Sensory systems	Extrinsic: bio-signals
			Intrinsic: inertial and piezoelectric sensors, potentiometers, encoders, load cells, strain gauges, etc
		Control methods	Finite-state machines, impedance based, adaptive, PID

measured by means of ink and paper, foot switches, treadmills, inertial measurements, and motion capture systems [6]. Being the last is the most popular method used for research in recent years, mainly for gait analysis on the sagittal plane.

In addition, some gait features have been identified to define other types of gait besides of level walking. For example, on stair ascent, the first point of contact with the ground has been defined as the forefoot (first metatarsal head), instead of the heel (as considered for level walking). Also, assessments made by Pinitlertsakun showed that the height of a step does not influence the range of motion of the knee, while the human body height does [22].

2.1.2. Kinetics

External torques over knee and ankle joints are frequently computed by solving the inverse dynamics problem. However, some authors have proposed alternative methods in order to estimate kinetic parameters during the gait cycle.

The study published by Wentink et al. [23] assessed kinetic gait parameters of transfemoral amputees using Inertial Measurement Units (IMUs) attached to the residual limb and the prosthesis, while other authors proposed the use of force transducers located directly under the transfemoral socket to obtain kinetic data of the knee joint, mostly to study deviations between normal and pathologic gaits [24–26]. However, in many studies the collected data showed similar values to those obtained with optoelectronic systems.

2.1.3. Surface electromyography

Myoelectric control has widely applied on upper limb prosthetic devices. The main uses of these signals on lower limb prosthetics are as an indicator of the movement intention on the early stages of level and slope gait, and on finite-state machines (FSM) as trigger signals for gait pattern detection on level, slope and stairs gait [2,23,27–29]. For example, Huang et al. characterized myoelectric patterns during level walking on stance and swing phases for lower limb prosthetic applications [30].

Transfemoral amputees and non-amputee volunteers have shown similar myoelectric activity patterns during level walking [8,12]. Au et al. [31] presented some interesting features on the use of surface myoelectric signals to control an active prosthesis for a transtibial device. Pressure and angle position sensors controlled parameters within two FSM for level and stair descent gaits. Myoelectric signals of the residual limb allowed the control system to switch between the FSMs.

One of the main challenges for the use of surface EMG is the data acquisition inside of the prosthetic socket. The quality of the surface EMG signal is highly sensitive to movement artifacts. Soft tissue on the residual limb could influence the skin-electrode contact inside of the socket. This issue has been addressed by Hefferman et al. [32] where four designs of instrumented transfemoral sockets were evaluated for different ADL, including level and stair walking. The best configuration for their experimental tests was an ischial containment socket with suction suspension. Such socket was assembled using wireless myoelectric electrodes on the inner walls over selected muscles. This design presented lower signal artifacts while their volunteers evaluated it as the most comfortable design.

An alternative solution to the use of surface EMG to characterize the gait cycle was reported by El-Sayed et al. [33], where the recognition of movement patterns was measured by Force Sensing Resistors (FSRs) and piezoelectric sensors on the anterior/posterior and distal/proximal walls of a transfemoral socket, respectively. The authors concluded that piezoelectric sensors were more accurate to detect medial and terminal swing phases, as well as standing and stair ascent, while the FSRs were more suitable to recognize stair ascent.

2.1.4. Metabolic cost

Metabolic cost is directly related with residual limb length and the kind of amputation (vascular or trauma). This index reflects the efficiency of the gait cycle. Common techniques to measure metabolic cost are indirect calorimetry, exhalation measurement, and heart rate monitoring (also known as Physiological Cost Index) [6].

Besides residual limb length, joint stiffness is directly related with the metabolic cost during normal and prosthetic gait. For example, in the studies reported by [34,35], authors concluded that a reduction on the metabolic cost of prosthetic gait could be reached with an active assistance during plantarflexion at the instant where the forefoot lifts up from the ground. According to Esquenazi and Talaty [6], the deceleration during terminal swing, heel strike, and forward progression are the most energetically demanding stages during level gait.

2.2. Biomechanical design: actuators

Actuators for semi-active prosthesis have evolved from pneumatic to hydraulic and rheological devices, while active prosthetics have been usually assembled with electro-mechanical actuation systems. The most common election in recent years for such servo-mechanisms in prosthetic applications include the assembly of electric actuators (commonly a motor and ball screw assembly) along with passive elements (springs or dampers) in order to reduce impact forces and to improve force control during gait assistance, while maintaining inertial parameters close to those of the biological limb. These types of mechanisms are known as Series Elastic Actuators (SEAs), whose terminology was first proposed by Pratt and Williamson in 1995 [36]. The most simplified model of a SEA is shown in Figure 1, where an electric motor is connected in series with a linear spring which transfers forces from the assembly to the load. An improved force control is achieved by measuring the strain on the spring coupled to the motor instead of measuring parameters directly on the electric motor. These actuators have beneficial properties for lower limb prosthetic applications such as low impedance, low friction, improved bandwidth, and high force-sensitiveness [37]. Figure 2 describes an example of the concept of a SEA on lower limb prosthetic applications.

For example, the work described by Eslamy et al. [38] showed experimental results where a single passive element (low stiffness spring) coupled in series to an active actuator was the most efficient assembly in comparison with a series elastic-damper and a parallel elastic-damper actuator. The tests performed by the authors were aimed for biomechanical applications.

2.3. Control systems

Specific control strategies have been applied for semi-active and active lower limb prosthetic control to assist the amputee during the gait cycle. The most common methods are FSMs and impedance control. One advantage of these strategies is that their mathematical models are not time-dependent, which is beneficial for prosthetic applications in contrast to other control algorithms such as tracking control [39]. Thus, changes in cadence during prosthetic gait would not affect the assistance for FSMs.

FSMs on prosthetic control can respond to external inputs, mainly from embedded sensors on the prosthetic structure to identify the states and transitions within phases of the gait cycle. The gains on control algorithms are modulated between states in order to actively adjust joint stiffness or regulate force on the actuators. In many cases, FSMs are combined with impedance control strategies in order to enhance the stability during prosthetic gait on different terrains, while other authors have applied algorithms such as PID or adaptive control [40].

Control algorithms on prosthetics require sensorial inputs to perform the gait cycle. While each prototype reported on the literature has proposed its own control method, some common sensor selection has been observed on this subject. The most common sensors employed to feedback information to the control system are position (potentiometers or encoders), force (force sensing resistors, strain gauges, transducers, and load cells), and inertial sensors (accelerometers).

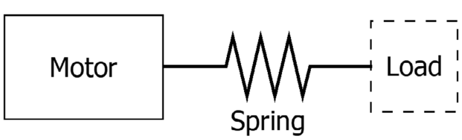

Figure 1. Simplified model of a Series Elastic Actuator proposed by Pratt and Williamson [36].

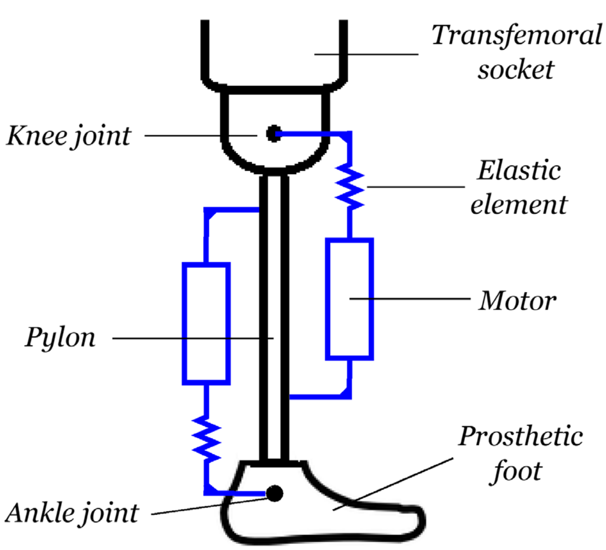

Figure 2. Schematic diagram of Series Elastic Actuators assembled on a transfemoral prosthesis for knee and ankle joints.

3. Active transfemoral prosthetics

Transfemoral prosthetic devices have been continuously developed in the search of the most natural prosthetic walk. In this section, a description of a brief selection of semi-active devices is presented along with some of the latest developments on active transfemoral prosthetics. Distinctive features of each device are described, as well as their actuation mechanisms, sensorial systems, and control methods. The final part of this section describes the main results of experimental assessments made to selected devices in order to evaluate the quality of their prosthetic assistance.

3.1. Semi-active

The main function of semi-active prostheses is to regulate joint stiffness by processing sensorial data during the gait cycle. These devices are usually restricted to only dissipate energy in order to enhance stability by delivering negative work to the joints. Some of the most outstanding semi-active devices are listed below.

(a) *Intelligent prosthesis (IP)*. The *Endolite's Intelligent Prosthesis Plus* commercialized by the Blatchford/ Endolite company was known as the first prosthetic knee system controlled by a microprocessor.

The joint velocity on the artificial knee was modulated during swing phase through an external device on the hand of the user to calibrate the level of assistance [41,42].

(b) *C-Leg*. The *C-Leg* is one of the most prescribed semi-active prosthetic knees to date. Developed by the Ottobock company, this device was the first prosthetic knee including a hydraulic damper controlled by a microprocessor. A hydraulic valve reacts in order to modulate gait velocity depending of the terrain using knee angle sensors and force transducers on the prosthetic leg [43,44]. The knee joint is provided with a mechanical locking system for standing. The damping parameters are adjusted manually by a trained prosthetist to optimize the quality of assistance [29]. According to the latest commercial brochure of Ottobock, the last generation of the C-Leg senses data from knee angle sensors and IMUs to measure position, velocity and acceleration of the prosthesis.

(c) *Rheo knee (RK)*. A microprocessor controlled prosthetic knee which is adaptable to different ground conditions using a magneto-rheological actuator This actuator was developed at the Massachusetts Institute of Technology (MIT) to act as a brake dissipating energy during stance phase [29]. The magneto-rheological fluid operates in shear-mode between concentric discs separated by a few microns functioning as rotors and stators inside of a chamber. In the presence of a magnetic field this actuator could reach up to 50 Nm of torque at slow speeds, while in the absence of a magnetic field the knee joint can move freely [45]. The control scheme for this prosthesis was based on an FSM for level walking. This device had embedded sensors in order to obtain force (axial and moment) data, angular position, and gait cadence for transitions within the state machine.

(d) *Sapporo Medical University*. This transfemoral prosthesis had a polycentric mechanism on the knee joint coupled to two actuators. A hydraulic actuator assisted during stance phase while a pneumatic actuator controlled by microprocessor assisted during swing phase. The control on the pneumatic actuator was able to adjust knee joint extension according to the walking speed [46].

(e) *University of California, Berkeley*. The operation of this prototype was based on a FSM controlling a hydraulic pump powered by an electric motor and a variable position valve. The pump transmitted active assistance during the stance phase, while the valve controlled joint stiffness during the swing phase. The FSM was able to determine if the amputee was standing, sitting or walking at level surfaces, slopes or stairs. This control scheme is based on predefined thresholds from these sensors. The level of assistance was modulated by using gait speed. The embedded sensors used were IMUs, encoders, pressure sensors and a force transducer [47,48].

3.2. Active

Active prosthetics are powered devices developed to perform more demanding gait activities (in comparison with level gait) such as ascending/descending slopes or stairs, as well as high impact activities such as sporting or even dealing with stumbles. Most of the active prosthetics are currently under continuous research as prototypes since many of their developments started in the early 2000s. This section describes the most representative active prostheses on this field along with some of their most outstanding features .

(a) *Waseda Leg*. The *Waseda Leg* was developed at the Waseda University in Japan on 1987 by Koganezawa et al. [49]. The WLP-7R (*Waseda Leg Prosthesis – type7 Refined*) was one of the first transfemoral prosthesis designed to supply active assistance, mainly for stair ascending and descending. Its working principle was based on a direct linkage between knee flexion and ankle dorsiflexion through a hydraulic circuit, as shown in Figure 3(a). The actuation mechanism was activated as the gait cycle advanced both for level walking and stair descent.

(b) *Power Knee (PK)*. Developed by the company Össur, this device was the first commercially available active transfemoral prosthesis. The actuation mechanism of the first generation was based on a linear actuator and a ball screw assembly as a SEA [50], as described in Figure 3(b). Assistance within a FSM was controlled with sensorial data using a load cell on the SEA and a potentiometer on the knee joint along with data of the intact limb [43,51]. According to Lambrecht and Kazerooni [47], one of the drawbacks of this device is that due to its geometry, only a 50% of male population could wear this device. This represented a limitation for the compatible prosthetic ankle-foot options. The second generation of the PK was driven by a harmonic drive motor, solving the size and weight limitations [52,53].

(c) *Vanderbilt University*. The active transfemoral prosthesis developed at the Vanderbilt University in the United States is one of the most developed active prototypes to date. Since 2006, three generations of this prototype have been developed. The first generation was driven by pneumatic actuators controlled by custom servo valves for knee and ankle joints. One of the aims of this design was to assist stair ascent and level walking. The instrumentation was composed by torque and angle sensors at each joint, as well as a load cell to measure sagittal and frontal interaction forces [54,55]. The control scheme was a finite-state-based impedance control, modeled as a series spring-damper mechanism. The gains of this controller were related to stiffness and damping gains on the model; which were modulated as the FSM performed transitions between gait phases from sensorial data. The second generation of this prosthesis replaced the actuation mechanism using a motor and ball-screw assembly for both joints, maintaining the same sensor and control schemes [56]. The work described by Varol et al. [57,58] reported the assessment of an algorithm for movement detection using the current sensor information of this prosthesis. The main goal was to enable the prosthesis to identify the scenario where the amputee requires assistance such as: standing, sitting, stair, and level walking. Also, some tests were performed with non-amputee volunteers wearing an 'able bodied adapter' to walk using the prosthetic leg instead of their biological one. The standing-to-level walking studies with amputee volunteers were performed on [59]. Surface EMG was included in 2010 to improve the impedance control scheme [60], as well as IMUs embedded to thigh, leg, and foot for stumble detection [61–63]. The third generation of this prototype included an additional state on its FSM for the swing phase (for a total of five states within the FSM) [64], while the actuation mechanism was replaced with electric motors and belt/chain transmissions [65] The sensing system on the last generation includes measurements of angular data on both joints and axial load in the shank [66]. The actuation scheme of this prosthesis is shown in Figure 3(c). The Vanderbilt Prosthesis has developments in collaboration with the Rehabilitation Institute of Chicago (RIC). One of the main contributions to this project was a myoelectric control scheme for transfemoral amputees with muscular reinnervation. This control mode allows the amputees to actuate each artificial joint independently using myoelectric signals recorded on the residual limb [67].

(d) *MIT Prosthesis*. An active transfemoral prosthesis was presented by the MIT on 2008. The actuation mechanism was an agonist-antagonist arrangement of SEAs reaching up to 100 Nm of torque for level walking. The control of this actuator was developed as a five-state FSM for level walking at the knee joint. The sensors on this device comprise an IMU, encoders and hall effect sensors. Authors claim that this device has a combination of quasi-passive and active actuation schemes [68]. A scheme of its working principle is shown in Figure 3(d). The joint stiffness was variable by using a mechanism which allows flexion and extension springs to couple independently to each actuator. The transitions between states within the FSM were achieved according to heel and toe contact with the ground, as well as knee angle [69]. One of the latest developments of this prototype assessed simulations of a clutch attached to the SEAs to improve dynamics and energy consumption [70].

(e) *Clarkson University*. The first approach of this prototype was conceptualized as an active transfemoral prosthesis with an energy recovery system acting on the swing phase [71]. A myoelectric algorithm of a prosthetic arm was used for the control of this device [72]. The first functional prototype was reported in 2012. The instrumentation was based on four surface-myoelectric electrodes inside of a suction transfemoral socket including a potentiometer and a uniaxial load cell on the knee joint. Inside of the shoe, pneumatic chambers recorded heel and toe forces. The actuation mechanism used for the knee joint was a ball screw assembly. The ankle joint was assembled with a passive commercial prosthetic foot. The main outcome of this device was the quality of the recorded myoelectric signals, which presented noise due to sweat and relative movement between the residual limb and the socket. The control system was based on a proportional plus feed-forward control law on the servo-actuator combined with myoelectric control. [73]. A scheme of the actuation mechanism is presented in Figure 3(e).

(f) *ANGELAA*. Since 2010, the Swiss Federal Institute of Technology in Zurich (ETHZ) developed a prototype of active transfemoral prosthesis called ANGELAA (Angle-dependent Elastic Actuator). The prosthetic assistance was accomplished using path tracking along with kinematic data of sensors on the contralateral leg, known as Complementary Limb Motion Estimation (CLME) [74]. During the design stage, the authors assessed possible joint geometries for a prosthetic knee to find the optimum stability values. While polycentric joints improve stability for passive transfemoral prosthetics, simulations performed by Pfeifer et al. [75] showed that an active transfemoral prosthesis with monocentric knee joint (a ball screw assembled to a slider-crank mechanism) can improve the angle-torque relation. The actuator mechanism was proposed as a *Series visco-elastic actuator*, where rubber cables allowed an improved joint stiffness even with the effects of hysteresis. A scheme of this working principle is presented in Figure 3(f). Besides CLME control, experimental tests were developed controlling this device with a FSM which divided gait cycle in four states using a control law based on a series spring-damper mechanism. The sensors used were encoders for knee joint angle, a load cell to measure knee moment and EMG recordings to estimate joint stiffness [76].

(g) *CYBERLEGs*. The Scuola Superiore Sant'Anna in Italy started the CYBERLEGs (Cybernetic Lower-Limb Cognitive Ortho-prosthesis) project in 2012. This prototype combines a lower limb orthosis with an active transfemoral prosthesis for vascular amputees. This device collected data from the biological leg to control the assistance of the prosthetic leg. On its alpha prototype, the knee joint was passive while the ankle joint was active. The early development of this prototype took an advantage of previous studies such as a pressure sole design [77,78] and an energy recovery mechanism to transfer energy from the prosthetic knee to the ankle by using springs and pulleys. The sensory system included IMUs along with the pressure soles [79,80]. In 2014, a case study of three amputee volunteers showed kinematic parameters during the swing phase close to the biological kinematics of level walking [81]. The control scheme was based on a FSM for standing and walking. Transitions between states were based on kinematic data of both the orthosis and the prosthesis [82]. A beta prototype was presented in 2015, where the prosthetic joints were both active to assist stair gait. The knee joint was powered by a SEA [83]; its actuation mechanism is depicted in Figure 3(g).

(h) *University of Rhode Island*. The FSM controlling this prototype included a methodology based on the Dempster–Shafer Theory. This theory allows the controller to detect transitions on the FSM with a greater accuracy compared to a sensor-only control system. The actuator was assembled as a slider-crank mechanism along with a ball screw. The angular position of the prosthetic knee and a load cell under the pyramid of the prosthesis provided information to the five-states on the FSM. Each state was controlled by an impedance model [84]. The mechanical model of the actuation assembly is shown in Figure 3(h).

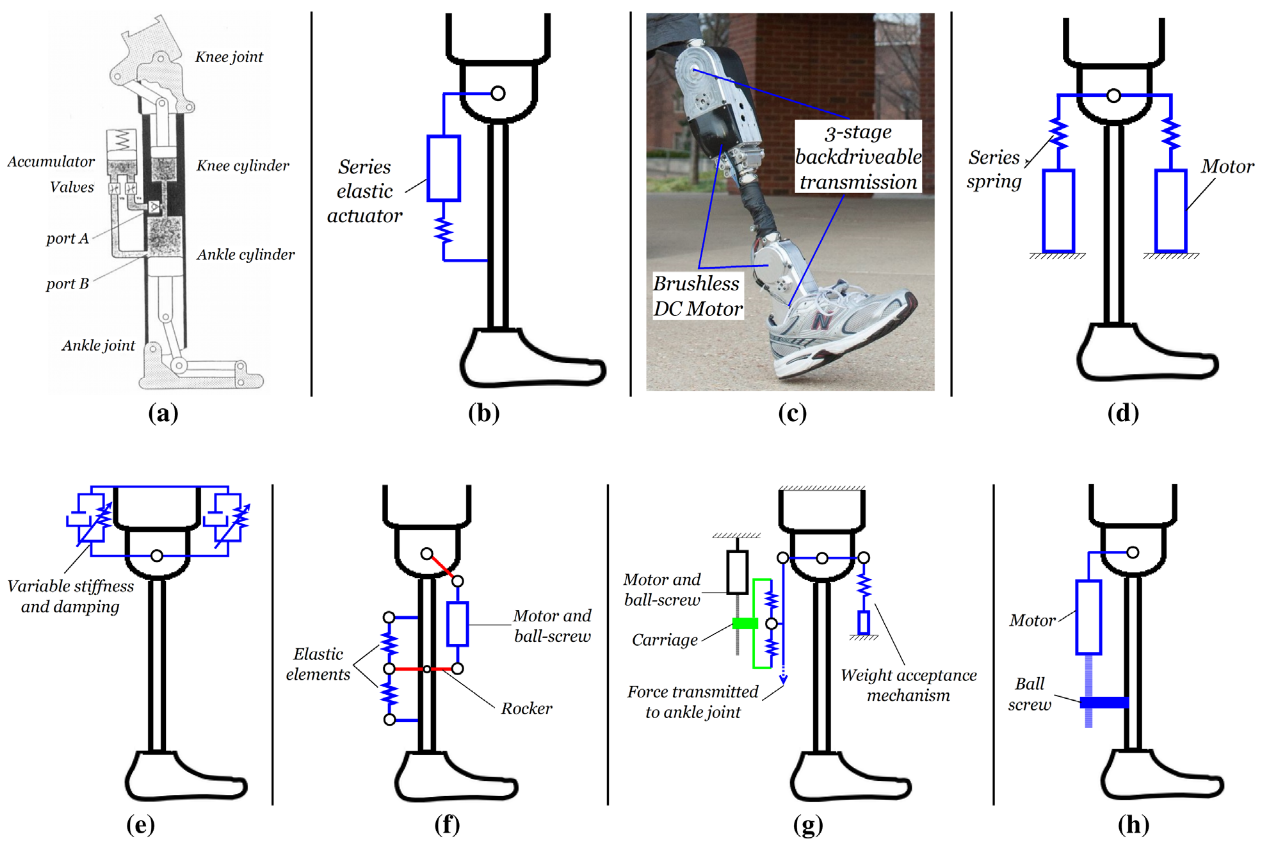

Figure 3. Actuation mechanisms on active transfemoral prosthetics. Working principles were defined based on cited publications. (a) Mechanism of the Waseda Leg [49] (image credit to Dr. Koichi Koganezawa), (b) working principle of the Power Knee [50], (c) third generation of the Vanderbilt University prosthesis [66] (image credit to Dr. Michael Goldfarb), (d) working principle of the antagonistic MIT prosthesis [85], (e) variable impedance mechanism on the active prosthesis of the Clarkson University [72], (f) conceptualization of the serial compliant elements as rubber cords on the ANGLEAA prosthesis [86], (g) schematic view of the knee system on the beta prototype of the CYBERLEGs project [83], (h) scheme of the actuation mechanism on the prosthesis of the University of Rhode Island [84].

3.3. Performance assessments on transfemoral prosthetics

The main goal of active prosthetic devices for lower limbs is to restore the quality of life (QOL) for amputees on ADL. The quality of assistance during level walking on active prosthetics for lower limbs has reached a close-to-natural gait, mostly on its kinematic and kinetic patterns. For example, the first experimental results of the Vanderbilt University prosthesis on an amputee volunteer showed improved prosthetic biomechanics on knee and ankle joints [4,87]. Quantitative performance results have been found between passive, semi-active, and active lower limb prosthetics, mostly through gait analysis and metabolic cost measurements. These kinds of assessments have shown the increase on the efficacy of lower limb active prosthetics during gait assistance. In this section, some of the many results obtained by researchers are described.

Buckley et al. [88] compared the assistance of the IP to a passive pneumatic prosthesis during the swing phase. The authors measured the metabolic cost in both devices and they found that the IP decreased the metabolic cost during level walking, even when the IP was heavier than the passive device.

Chin et al. [3] carried out an estimation of the metabolic cost during level walking at four different speeds. The experiment was performed between the IP, the C-Leg, and a control group of non-amputees. The results showed an increased metabolic cost for both prostheses between 50 and 60% in contrast with the control group. However, the authors did not find significant differences on the biomechanics between both prostheses.

The C-Leg and the PK were also compared in [89,90]. The authors reported similar efficacy for chair activities and slope ascent. The PK showed improved results in gait ascent due to its active design.

The performance of the second generations of the PK and the RK prostheses was measured during level, slope, and stair walking in 2012. Overall, the PK showed a close-to-natural gait kinematics during stair and slope ascent in comparison to a control group. However, the external moments at hip joint on the amputee limb were greater than those of the control group. The ranges of motion and the moments at the artificial knee joint were smaller for the PK in comparison with the RK, proving the benefits of active lower limb prosthetics in comparison with semi-active prosthetics during slopes and stairs ascent tests [91].

The MIT prosthesis was assessed against the C-Leg during normal gait. The MIT prosthesis shown a reduction of the metabolic cost of around 6.8% in comparison with the C-Leg. According to the authors, this was the first time that an active knee and ankle-foot prosthesis showed a reduction on the metabolic cost in comparison with a passive transfemoral prosthesis [85,92].

In 2015, the transfemoral active prosthesis of the Vanderbilt University was compared against a passive device in order to compare the metabolic cost during stair ascent. The results reported by Ledoux et al. [65] showed that the active device reduced the metabolic cost by around 30%.

4. Discussion and conclusions

Researchers on the field of biomechanics and prosthetic design are currently studying three current trends: the analysis of different gait cycles and their transitions, actuator design and optimization, and control systems.

The reviewed studies related with the gait analysis on stair ascent/descent have shown significant variations on the experimental conditions during their tests. Important aspects such as methodology and laboratory or volunteer conditions can make the results difficult to compare. Standardized tests for stair gait of amputee and non-amputee volunteers can lead to an improvement and collective characterization of the different gait cycles; i.e.: standard dimensions and number of steps, gait symmetry measurements, standard methods for metabolic cost measurements, dynamic EMG methodologies for in-socket detection, etc.

As foreseen, some of the latest prototypes have already reached improved kinematics for normal gait; even close to biological levels. From a mechanical design perspective, one of the future challenges in this field is the optimization and the energy efficiency for actuation mechanisms and power supplies to develop more autonomous lower limb active prosthetics. Also, lightweight designs in relation to the weight of the loss limb are important to minimize metabolic cost. The progress on novel actuator designs (like the SEAs) is also an essential part in order to achieve these objectives.

The active myoelectric control for lower limb prosthetics is still a challenge. Actual solutions in this field point to focus on volitional control of prosthesis during transitions on the different types of gait cycles combining signals of embedded sensors in the mechanical structure and the prosthetic socket to feed information to the FSMs or to the impedance control strategies. Full volitional control for active lower limb prosthetics is one of the main objectives for researches in the field of prosthetic design, mostly for transfemoral amputees who need to increase their walking stability during ADL due to the absence of biological knee and ankle joints.

While most of the reviewed publications on active transfemoral prosthesis are prototypes under development, and since there are experimental tests with volunteers wearing these devices, there is a lack of specifications related with the mechanical design (i.e. fatigue life due to the workloads during walking), even when at least one international standard is available for structural testing on the design of lower limb prosthetics to ensure user safety (ISO 10328).

The design of transfemoral active prosthesis for amputees continues on the way to develop devices with improved capabilities in order to assist more ADL every time. Lower metabolic costs, optimum and variable gait velocity, and symmetric gait are some of the objectives in engineering to provide a more stable assistance, especially for more demanding activities such as stair gait or running. From a biomechanical and medical perspective, a more human-like prosthetic gait can reduce the risk of injuries due to the long use of passive devices. Based on the reviewed devices in this study, we conclude that in a near future, transfemoral active prosthetic devices must be able to restore the QOL for lower limb amputees.

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