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Biomechanical Considerations in the Design of Lower Limb Exoskeletons

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Abstract— **This paper presents an analysis of the human biomechanical considerations related to the development of lower limb exoskeletons. Factors such as kinematic alignment and compatibility, joint range of motion, maximum torque, and joint bandwidth are discussed in the framework of a review of the design specifications for exoskeleton prototypes discussed in the literature. From this analysis, we discuss major gaps in the research related to the topic and how those might be filled.**

Index terms – exoskeleton, robotic, design, lower extremities

I. INTRODUCTION

Exoskeletons are electromechanical devices that are worn by a human operator and designed to increase the physical performance of the wearer. This performance increase might include increased load carrying capacity, lower metabolic expenditure, or running at faster speeds or for longer distances, [1-3]. Because of the close interaction between the wearer and the exoskeleton, these devices must be mechanically compatible with human anatomy, able to safely move in concert with the wearer without obstructing or resisting movement [1, 4, 5].

There is prior work relating to lower limb exoskeletons that spans the past four decades [6, 7], and research on the topic has been more active in the recent years [3, 6-8]. However, there has yet to be any substantial successes published in the literature in which a lower-limb exoskeleton has increased the performance of an able-bodied wearer. One of the major challenges in exoskeleton research is a lack of understanding of the underlying mechanisms that are responsible for control of movement in humans and how those interact with a robotic device in parallel with the wearer [7]. As a result, it is impossible to know how various design, sensing, and control choices should be made in order to maximize the performance of the wearer while minimizing the interference with the wearer's preferred movement strategies.

This paper reviews the current literature and discusses the design and performance specifications of a number of popular lower limb exoskeleton devices, particularly as they relate to human biomechanical considerations. The exoskeletons we consider in this review are those that act in parallel to the human body of the wearer – the most common design configuration, as opposed to other exoskeletons that act in series [8]. For this reason, the device kinematics must be compliant with that of the human limb [1]. Additionally, not all devices span the entire leg. Some designs span just one joint such as the ankle or knee, while others span two or all three. Additionally, due to the different target applications of the devices (e.g. load carrying, running, etc.), most of the hardware realizations are application dependent and therefore differ in the choice of the mechanics, actuation, control, and other design parameters [1].

We begin the remainder of this paper with a summary of the designs of some of the most widely known exoskeleton prototypes, paying particular attention to those in which quantitative justification is provided for the design choices. After the relevant design parameters are presented in table form, we discuss the design specifications and parameters, going joint by joint to analyze considerations such as degrees of freedom, kinematic alignment, range of motion, torque, speed, and others. Finally, we discuss how these results inform future research in the area in order to help bring the successful implementation of these devices to fruition.

II. CURRENT EXOSKELETON DESIGNS

The primary aim for this review is to summarize the current state of the science that lies behind the design of exoskeletons. We present the biomechanical aspects and challenges to consider when designing a lower limb exoskeleton that acts in parallel to the limb. In doing so, we also review how those challenges were dealt with or addressed in some exoskeleton prototypes available in the literature. Some of these exoskeletons are for augmentation of the able-bodied wearer physical abilities [2, 4, 9, 10], while others are designed to assist the wearer in the case of a diminished functionality to stand and locomote [11-14]. Also, with the exception of the MIT Knee Exoskeleton [9] that was designed for running, all the others devices are solely or primarily focused on human walking [2, 4, 10-14]. In this review article, the HAL [15, 16] and the Sarcos [6] were not included in the biomechanical considerations because of the limited amount of information reported in the literature about these two devices.

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TABLE I. MECHANICAL FACTORS RELATING TO DESIGN OF LOWER LIMB EXOSKELETONS

			Augmenting Exos				***Assistive Exos***			
Bio-mechanical Properties	**Joints**	***Values for Biological Lower Limb [1, 17-19]***	***BLEEX (Univ. of California) [4, 20, 21]***	***MIT Exoskeleton (MIT) [2, 5, 22]***	***MIT Knee Exoskeleton (MIT) [9]***	***NTU-LEE (Nanyang Technological Univ.) [10]***	***DGO/Lokomat (Hocoma, Switzerland) [11, 23]***	***LOPES (Univ. of Twente) [12, 24, 25]***	***KNEXO (Brussels University) [13]***	***ALEX (Univ. of Delaware) [14]***
DOF	***Hip***	3	3	3	N/A	2	1	2	1 (un-actuated)	2
	Knee	2	1	1	1	1	1	1	1	1
	Ankle	1 (+1)[a]	3	2	N/A	2	N/A	N/A	N/A	1
ROM [deg]	***Hip***	140/15 (F/E)[b]; 40/30-35 (Ad/Abd)[c]; 15-30/60 (Int/Ext)[d]	121/10 (F/E)[b]; 16/16 (Ad/Abd)[c]; 35/35 (Int/Ext)[d]	45/20 (F/E)[a]	N/A	60/20 (F/E)[b]	N/A	60/30 (F/E)[b]; 15/15 (Ad/Abd)[c]	N/A	N/A
	Knee	120-140/0-10 (F/E)[b]; 10-15/30-50 (Int/Ext)[d]	121/0	90/0	100/0	120/0	N/A	90/0	90/0	N/A
	Ankle	40-50/20 (F/E)[b]; 30-35/15-20 (Inv/Ev)[a]	45/45 (F/E)[b]; 20/20 (Inv/Ev)[a]	15/15 (F/E)[a]	N/A	30/20 (F/E)[b]	N/A	N/A	N/A	N/A
Torque [Nm]	***Hip***	140/120 (F/E)[b] [Walking] 40-80 [Running]	-150:120 (F/E)[b]	130	N/A	118 (F/E)[b]	50 (280 max) (F/E)[b]	65 (F/E)[b]; 30 (Ad/Abd)[c]	N/A	100 (F/E)[b]
	Knee	50/140 [Walking] 125-273 [Running]	-100:120	50	~135	118 (F/E)[b]	30 (160 max) (F/E)[b]	65	70	100 (F/E)[b]
	Ankle	165 (E)[b] [Walking] 180-240 [Running]	-200:150 (F/E)[b]	90	N/A	118 (F/E)[b]	N/A	N/A	N/A	N/A
Velocity [rad/s]	***Hip***	N/A	N/A	4	N/A	N/A	N/A	1 (Ad/Abd)[c]; 2 (F/E)[b]	N/A	N/A
	Knee	N/A	N/A	N/A	10.5 (F/E)[a]	N/A	N/A	5 (F/E)[b]	10 (F/E)[b]	N/A
	Ankle	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
Frequency [Hz]	***Hip***	N/A	N/A	N/A	N/A	N/A	≥1	4 (full force range); 12 (small forces)	N/A	N/A
	Knee	N/A	N/A	N/A	N/A	N/A	≥1	4 (full force range); 12 (small forces)	3.5-4.5	N/A
	Ankle	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A

a. The foot has 3 DOF and is characterized by Inversion/Eversion

b. Flexion/Extension

c. Adduction/Abduction

d. Internal/External

The critical biomechanical factors to consider in the design of an exoskeleton, such as degrees-of-freedom (DOFs), range-of-motion (ROM), joint torque requirements, joint rotational velocity, and joint angular bandwidth are reported in Table I, which also lists the values that pertain to the biomechanical properties of the biological limb to allow easy comparison with the proposed solution. Other factors that pertain more to the type of actuation, weight distribution/inertia of the exoskeletons, and physical interfacing with the user body are reported in Table II.

A. Kinematic Considerations

The exoskeletons considered in this review act in parallel to the human body. An exoskeleton acting in parallel to the limb means that the two systems (limb and device) will have to move together at any given time without restricting each other's motion [1, 21]. This concept is presented in [1] as kinematic compliance (in the ideal case), meaning that the exoskeleton mechanics will comply with that of the limb and therefore will not interfere with its natural motion. Ideally, the joint center of the exoskeleton should be aligned with that of the biological limb, have enough DOF to allow free, un-restricted motion of the limb, and be able to provide torques at the joints that are compatible with those of the human body [1, 21]. These mechanical requirements for exoskeleton design implicitly assume we have a good characterization of the human limb biomechanics. However, the non-ideal mechanics of the joints and segments makes the design task challenging [21], in addition to the fact that the mechanics of gait change substantially from one user to the next. For the purpose of this review, the biomechanical properties considered that relate to the kinematics of the lower limb include a joint's DOFs, ROM, torques, rotational velocity, and bandwidth (Table I).

1) Degrees of freedom

a) Hip

The hip joint has three DOFs that are all rotations and therefore is considered to be a ball-and-socket joint [19]. The allowed motions at this joint are flexion/extension, adduction/abduction, and internal/external rotations (Table I).

In Table I, all the exoskeletons that include the hip joint include at least flexion/extension, which is the primary DOF used in locomotion [21]. This DOF doesn't seem to present a major design challenge as the limb lies in a plane containing the axis of rotation for this DOF. This is not the case for the adduction/abduction rotations for which the limb is offset from the joint center. Therefore during an adduction/abduction rotation there is a relative movement between the two limbs. In [5] (MIT exoskeleton) and [4] (BLEEX), the authors acknowledge that they did not take this issue into account in their original designs. Each of these two groups proposed a different solution to account for change of limb length during hip adduction/abduction rotations. For the MIT Exoskeleton, they opted for a cam mechanism (see [5, 26] for more details), while in the BLEEX, they positioned the center of rotation for this DOF in the rear part of the hip joint mechanism. This solution allows the center for flexion/extension rotations to move accordingly and stay aligned with the biological hip joint center. According to the groups, both solutions improved the capacity of the exoskeleton to be kinematically compliant with the this DOF of the hip joint, but no quantitative data is given in support of their conclusions regarding this issue. Both the LOPES [12] and the ALEX [14] implemented a solution for the hip abduction analogous to the one chosen in the BLEEX. However, no specific justification was provided about the design of the realization of adduction/abduction DOF for the hip in [10, 12, 14].

The last DOF of the hip is the internal/external rotation, which is a rotation along the longitudinal axis of the limb. This rotation suffers the same issues of the adduction/abduction as the limb is offset from the actual joint center. The combination of this offset and because the exoskeleton is external to the biological limb causes a relative motion between the two systems. The MIT exoskeleton [26] and the BLEEX [4, 21] were the only ones that included this DOF and they presented their solution in their publications. In the MIT exoskeleton, this rotation is permitted by placing a rotator joint above the knee, in the upper leg shaft [22]. No quantitative data to support the efficacy of this solution is provided, therefore we could not conclude whether or not this solution makes the system kinematic compliant for this DOF. The first solution implemented in the BLEEX for this problem was an axis that was located just above each leg [4], which is functionally similar to the solution used in the MIT exoskeleton. This solution, however, did not work well for the BLEEX, particularly under load-carrying conditions in which the moment on the un-actuated DOF increased. This aspect of this solution induced the designers to adopt a different approach. The two axes (left and right) were replaced by a single axis located posterior to the wearer and in the middle of the hip joint system. This axis allows the left and right parts of the whole joint to rotate internally/externally with respect to each other (see [4]) and therefore the axis is now able to follow the leg motion during these rotations. However, other than the assurance from the authors that this solution worked, we were not able to find any quantitative evaluation of the performance of this proposed solution.

b) Knee

The knee joint has two rotational DOF and is considered a condyloid joint [19]. The motions at this joint are flexion/extension and internal/external rotations (Table I). The knee joint is, however, often reduced to one DOF due to the very limited internal/external rotations. As reported in Table I, for all designs considered in this review, the knee joint is designed with one DOF: flexion/extension in the sagittal plane.

However, in [12, 25] (LOPES) the authors report that other knee motions are left free (un-constrained) to avoid any external force/torque applied to the knee joint. It is known that there is a motion in the frontal and coronal planes of motion, but they are not often reported [13]. From a design standpoint, this is an aspect to consider and is particularly important for load-carrying conditions. In those conditions, the load may add stress on the body by exerting forces and torques that were not accounted for in the original design.

c) Ankle

The ankle joint is considered to act as a hinge joint with one DOF that allows rotations in the sagittal plane (flexion/extension) (Table I). Any actuation applied to the ankle of an exoskeleton is applied to this DOF. However, because the ankle joint is connected to the ground via the foot, the complex structure and internal DOFs of the foot must be considered [19, 21]. The foot of the device is usually constructed to allow inversion/eversion, as reported in Table I. Both BLEEX and the more recent version of the MIT Exoskeleton have included this DOF in their design [4, 22]. According to [4], the BLEEX ankle joint also includes an internal/external rotation DOF, a mention to why is given in [21] but without quantitative data regarding the efficacy of this solution. In the BLEEX, the two axes of inversion/eversion and internal/external rotations were located laterally to the ankle joint (see [4] for details). The authors state that this choice was made to keep the design simple.

2) Range of motion

The ROM requirements of a joint can depend on the application. For instance, in rehabilitation the ROM for a particular DOF might be based solely on ROM observed in able-bodied subjects during the performance of a particular task. If a person is being treated to regain his/her ability to walk, then the ROM of an assistive exoskeleton could simply be the range for normal walking, plus a bit extra as a safety margin. As shown in [4], the ROMs for walking are actually less than the maximum ranges for the average military male [4], which are less than the minimum ROM that are reported in [17] for each joint DOF.

In Table I, we report values of ROM for each joint and each DOF that we were able to gather from the literature reviewed in preparing this manuscript [1, 17, 18]. Each of prototypes included in this review adopted ROMs for each of the allowed DOF that included the required ROM of the biological limb (see Table I). However, those ranges can differ depending on the particular application intended for the exoskeleton. For exoskeletons designed for walking, the ROMs were always larger than ROM reported for over ground walking [4, 13, 17, 22]. Therefore, those ROM were large enough to avoid any restriction to limb movement during locomotion. The MIT knee exoskeleton, having been designed for running purposes, had a knee flexion ROM of ~100 deg. This ROM is compatible with the physiological ROM of the knee reported in Table I [1, 17].

B. Joint Torques

In addition to kinematic compliance, an exoskeleton is meant to assist the operator during locomotion. This is done by applying torques either passively [2, 5, 9] or actively [4, 10, 12, 13] to the joints of interest at the right time, in the right direction, and of the appropriate intensity. In this subsection we cover primarily the intensity aspect as it pertains more to the design of the mechanical aspects of the system than the others, which are related to the problem of control.

It is widely accepted that walking and running activities are characterized by motion of the legs, and the largest joint powers are observed in the sagittal plane [9, 13, 21]. All of the prototypes presented in this review provide assistance primarily to the flexion/extension DOFs, and therefore in the sagittal plane. However, in [4] the authors pointed out that the hip joint requires the largest power in the adduction/abduction rotation than any other non-flexing/extending rotations. This explains why the BLEEX, the MIT Exoskeleton, and LOPES included some assistance to this hip DOF [4, 12, 22]. In addition, the designers of the BLEEX and the MIT Exoskeleton (the later prototype), included some support at the ankle inversion/eversion by including a spring system at this joint which is particularly important during load-carrying conditions DOF [2, 4].

Most authors seem to agree that clinical gait analysis (CGA) data sources are a good start for the initial design of the actuation to be used in their prototypes [4]. These data sets provide average locomotion data to base the design, in consideration that locomotion data is variable across subjects [4, 13]. Therefore, using average data helps in the design of a system that addresses general movement characteristics across a wide range of subjects. However, Beyl in his recent publication [13] makes a remark about the large variability observed in gait data and cautions designers of actuated exoskeletons be careful in the interpretation of CGA data and in the formulation of design recommendations based on those data.

Joint torque data is used to determine the required characteristics for the actuation to be applied at each assisted joint. The intensity of the joint moments fluctuates within the gait cycle [4, 9, 13, 27], and therefore in most cases designers use maximum values (peaks) as requirements for the sizing of their actuators [4, 9, 13, 25], However, in [2] the authors used optimization methods and models of human motion to estimate the required torques for their assistive passive systems.

C. Motion Velocity and Bandwidth Considerations

The kinematic (DOF and ROM) and kinetic (joint torques) characteristics of lower limb biomechanics represent the static characteristics of movements. However, gait is a dynamic task and therefore is characterized by velocity and frequency bandwidth of the movements. To our knowledge, speed and bandwidth considerations were only presented in [9, 11-13]. These aspects of motions are particularly important in the torque generation of the exoskeleton because actuators don’t act instantaneously (they have some dynamic behavior) and therefore the designer needs to chose an actuation system able to generate not only the required joint torques intensities, but with the necessary speed and adequate frequency response. Therefore, joint rotational speed and the amplitude and force bandwidth must also be considered for the specific application of the exoskeleton.

D. Considerations on Other Design Factors

The design factors covered in the previous parts of this review are closely related to kinematic compliance with the biological limb, to the ability to apply the correct joint torques, and to the dynamic behavior of the system in motion. These aspects are important in the design of lower limb exoskeleton but not the only ones. There are other aspects that designers need to consider for the final fabrication of a prototype. These are more related to whether the structure is anthropomorphic, the type of actuation to use to assist during the motion, the weight and the inertia of the device, and how it will be interfaced with the user. All those other design aspects are listed in Table II.

A device is considered anthropomorphic when the elements constituting the exoskeleton frame are sized following the proportions among the corresponding segments in the limb assisted. All the exoskeletons presented in this review, except for the BLEEX and ALEX, have been designed by sizing their components to human proportions. The BLEEX is considered almost- or pseudo-anthropomorphic [21] in the sense that the exoskeleton is only connected to the body at the torso and the feet, and in between the system doesn’t have to be constructed exactly as the human body as long as there is no interference with the motion of the leg (kinematic compliance) [1, 21]. This type of solution is analogous to what in [1] is termed “end-point based”. According to the authors of [21] this approach allows some relative motion without restricting the body helps to account for non-ideal joints such as the knee. This joint is not a rotary joint but is characterized by some sliding. For the ALEX system we could not find any mention about being anthropomorphic or not. For all the others that confirm to have designed the device following human proportions, only for the MIT Exoskeleton and for the KNEXO the sources of the human factor data were provided [5, 13].

One aspect that will have an impact on the amount of energy required to power the exoskeleton system is the choice of the actuators. In the examples included here, actuation choices included passive, quasi-passive or active. With quasi-

TABLE II. OTHER DESIGN ASPECTS OF EXOSKELETON DESIGNS

	Augment				*Assist*			
Exoskeleton Name	***BLEEX (Univ. of California) [4, 20, 21]***	***MIT Exoskeleton (MIT) [2, 5, 22]***	***MIT Knee Exoskeleton (MIT) [9]***	***NTU-LEE (Nanyang Technological Univ.) [10]***	***DGO/Lokomat (Hocoma, Switzerland) [11, 23]***	***LOPES (Univ. of Twente) [12, 24, 25]***	***KNEXO (Brussels University) [13]***	***ALEX (Univ. of Delaware) [14]***
Anthropomorphic	Quasi	Yes	Yes	Yes	Yes	Yes	Yes	N/A
Active/Passive Actuation	Active/ Passive	Quasi-Passive	Quasi-Passive	Active	Active	Active	Active/ Free	Active/ Passive
Exoskeleton Weight/Inertia	Mass/Inertia similar to that of human body.	Distal weight was minimized.	Total weight minimized. Most weight was at the joint to increase strength for loads and bending moments.	N/A	Total weight is 21 kg.	3 kg (joints included) 6 kg for whole leg.	N/A	N/A
Interface	Connections at torso and feet.	Shoulder straps, waist belt, thigh cuff, and shoe connection.	Custom-fitted knee braces with hard cuff at back of leg and velcro straps.	Inner exoskeleton secured with straps.	Soft pads and wide straps were used to connect the DGO to the limbs and to prevent skin pressure sores	N/A	Rigid cuffs in different sizes. Relative position is adjustable.	Torque/force sensors are placed between leg segments and exoskeleton.

passive we intend actuation systems that provide forces/torques passively, but the time the passive element is engaged and characteristics of the passive element are actively controlled as described in [9]. Purely passive systems are the most economical and are based on the research work done on passive walkers that can walk autonomously on a slight slope to use gravity as a way to power the system to compensate for minimal energy losses. However, current designs have yet to prove this choice performs well [5, 9, 28]. Power requirements represent a difficult challenge for augmenting exoskeletons as they are supposed to be autonomous devices and therefore carry their own power supply. On the contrary, rehabilitation (assistive) devices can be externally powered as they are normally used in a room setting and over a treadmill. Also, the active choice for actuation is justified for rehabilitation settings such as for the Lokomat, LOPES, and KNEXO [11, 13, 25]. These devices are designed to guide the motion of the lower limbs to follow normal gait patterns; a passive device will not be able to do so. The NTU-LEE is the only one in this review that is purely active [10], while the BLEEX, and the ALEX opted to provide active assistance only in the sagittal plane. However, the ankle joint in the ALEX is not actively assisted [14].

The information about the weight and inertia properties of the exoskeletons in this review were chosen following different criteria. The BLEEX tried to have weight/inertia of the segments similar to that of humans because of the exoskeleton acting in parallel to the body [4]. In the MIT exoskeleton, the weight of the lower limbs was minimized to reduce metabolic consumption as studies have shown that additional load increases metabolic energy consumption as the load becomes more distal [5]. For the MIT Knee Exoskeleton, because this device was designed to assist running, the overall weight was minimized as the exoskeleton was carried by the wearer. Also, the joint was designed to withstand the loads and the bending moments that this required, which made the device heavy at the joint [9]. For the LOPES, the authors only reported the values without providing comment about the inertia and weight of the device [25].

Lastly, in Table III, we report how the exoskeletons were connected to the body. Interfacing the exoskeleton with the human body is difficult due to the compliant nature of the flesh. The compliance of this layer makes the tracking of leg motion difficult , and especially the transmission of forces and torques to the body. Additionally, another aspect related to the human-robot interface to consider is the occurrence of skin pressure sores that should be prevented as stated in [11]. The use of straps appeared to be the most common way to secure the device to the leg [9-11, 22]. For the KNEXO, they reported the use of rigid cuffs with the relative positions adjustable to fit subjects [13], while the Locomat uses soft pads in addition to straps [11]. The BLEEX and the ALEX only gave minimal information about the interface (see Table II).

III. DISCUSSION

Research on exoskeletons has been quite active in the past four decades [6, 7]. Prototypes have been proposed to augment able-bodied users' physical performance [2, 4, 9, 10, 15, 29] or to assist human motion for rehabilitation purposes [11-14, 16]. The solutions proposed until now are very different from each other in terms of number of assisted joints and number of joint DOFs (see Table I). Also, we have seen in the previous section of this review how the solution to a kinematic compliance issue can be resolved very differently as in the case of internal/external rotations at the hip joint between the BLEEX [4] and the MIT Exoskeleton [22]. The designers of these devices provided some justifications to support the choice made, however, no quantitative data was presented to support one design choice versus the other [5, 21, 22] and therefore it is difficult to determine how effective those solutions are. Others don't provide any information about the design criteria used in their prototypes [15, 30]. In general, we find that quantitative evaluation of the effectiveness attributed to design choices is lacking and it makes the evaluation of those solutions difficult.

This review was meant to investigate the biomechanical aspects that are critical in the design of lower limb exoskeletons and review how the information about current prototypes can help in proposing some recommendations for design criteria for lower limb exoskeletons. However, it is evident from this review the lack of quantitative evaluation of the efficacy for the majority of lower limb exoskeleton solutions developed. It is hard to conclude how effective those solutions are and to formulate any design recommendation based on their solutions. Therefore, it is evident that further the knowledge about the science related to the design of exoskeletons is needed, particularly the mechanical aspects.

Among the many issues related to the design of a lower limb exoskeleton, perhaps the most challenging is the variability and uncertainty about the biomechanics of the human body and the specific wearer.

In the previous review section we briefly mention that in [13], the author warns designers about interpreting gait analysis data as a means to define requirements and recommendations for the design of actuated exoskeletons. We agree that designers have to be careful when dealing with the design of exoskeletons and using data from human motion, but most importantly, we feel that there is a need for a more formal testing and experiments to test how the human body interacts with different design solutions to try to learn more about the human-robot physical interaction that is one of the key factor in the success of an exoskeleton design. This last aspect is particularly true from the mechanical considerations we covered in this review. In the future we therefore hope to see more basic science research on human-exoskeleton interaction and greater quantification of the effectiveness of exoskeleton solutions.

REFERENCES

[1] J. L. Pons, *Wearable Robots: Biomechatronic Exoskeletons*, First ed. Chichester, UK.: John Wiley & Sons, Ltd., 2008.

[2] C. J. Walsh, K. Endo, and H. Herr, "A quasi-passive leg exoskeleton for load-carrying augmentation," *International Journal of Humanoid Robotics,* vol. 4, pp. 487-506, 2007.

[3] E. Guizzo and H. Goldstein, "The Rise of the Body Bots," in *IEEE Spectrum*. vol. 42, 2005, pp. 42-48.

[4] A. B. Zoss, H. Kazerooni, and A. Chu, "Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX)," *IEEE/ASME Transactions on Mechatronics,* vol. 11, pp. 128-138, Apr 2006.

[5] A. Valiente, "Design of a Quasi-Passive Parallel Leg Exoskeleton to Augment Load Carrying for Walking," in *Dept. Mech. Eng.*, M. S. thesis, Ed. MIT: Boston, MA, 2005.

[6] A. M. Dollar and H. Herr, "Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art," *IEEE Transactions on Robotics,* pp. 144-158, Feb 2008.

[7] D. P. Ferris, G. S. Sawicki, and M. A. Daley, "A Physiologist's Perspective on Robotic Exoskeletons for Human Locomotion," *Int J HR,* vol. 4, pp. 507-528, Sep 2007.

[8] H. Herr, "Exoskeletons and Orthoses: Classification, Design Challenges and Future Directions," *J Neuroeng Rehabil,* vol. 6, Jun 2009.

[9] A. M. Dollar and H. Herr, "Design of a quasi-passive knee exoskeleton to assist running," in *Proc IEEE/RSJ Int Conf Intell Rob Syst*, Nice, FR, 2008, pp. 747-754.

[10] X. Liu and K. H. Low, "Development and preliminary study of the NTU lower extremity exoskeleton," in *Proc IEEE Conf Cyber Intell Syst*, Singapore, 2004, pp. 1243-1247.

[11] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill training of paraplegic patients using a robotic orthosis," *J Rehabil Res Dev,* vol. 37, pp. 693-700, Nov-Dec 2000.

[12] H. van der Kooij, J. Veneman, and R. Ekkelenkamp, "Design of a compliantly actuated exo-skeleton for an impedance controlled gait trainer robot," in *Conf Proc IEEE Eng Med Biol Soc*, New York, NY, 2006, pp. 189-93.

[13] P. Beyl, "Design and control of a knee exoskeleton powered by pleated pneumatic artificial muscles for robot-assisted gait rehabilitation," in *Mechanical Engineering*, P. D. dissertation, Ed. Brussels University, 2010.

[14] S. K. Banala, S. K. Agrawal, and J. P. Scholz, "Active Leg Exoskeleton (ALEX) for Gait Rehabilitation of Motor-Impaired Patients," in *Proc IEEE Int Conf Rehab Robot*, Noordwijk, NL, 2007, pp. 401-407.

[15] S. Lee and Y. Sankai, "Power assist control for walking aid with HAL-3 based on EMG and impedance adjustment around knee joint," in *Proc IEEE/RSJ Int Conf Intell Rob Syst*, Lausanne, CH, 2002, pp. 1499-1504.

[16] H. Kawamoto and Y. Sankai, "Comfortable power assist control method for walking aid by HAL-3," in *Proc IEEE Int Conf Syst Man Cyber*, Hammamet, TN, 2002.

[17] H. P. Crowell, III, "Human Engineering Design Guidelines for a Powered, Full Body Exoskeleton," U.S. Army Res Lab, Aberdeen Proving Ground, MD ARL-TN-60, Jul 1995.

[18] H. P. Crowell, III, A. C. Boynton, and M. Mungiole, "Exoskeleton Power and Torque Requirements Based on Human Biomechanics," U.S. Army Res Lab, Aberdeen Proving Ground, MD ARL-TN-2764, Nov 2002.

[19] J. Hamill and K. M. Knutzen, *Biomechanical Basis of Human Movement*, Second ed. Baltimore, MD: Lippincott Willims & Wilkins, 2003.

[20] A. Chu, H. Kazerooni, and A. Zoss, "On the Biomimetic Design of the Berkeley Lower Extremity Exoskeleton (BLEEX)," in *Proc IEEE Int Conf Robot Aut*, Barcelona, ES, 2005, pp. 4345-4352.

[21] H. Kazerooni and R. Steger, "The Berkeley Lower Extremity Exoskeleton," *Journal of Dynamic Systems, Measurement, and Control,* vol. 128, pp. 14-25, Mar 2006.

[22] C. J. Walsh, "Biomimetic Design of an Under-Actuated Leg Exoskeleton For Load-Carrying Augmentation," in *Mechanical Engineering*. vol. . P. D. dissetation, Ed. MIT: Boston, MA, 2006.

[23] G. Colombo, M. Wirz, and V. Dietz, "Driven gait orthosis for improvement of locomotor training in paraplegic patients," *Spinal Cord,* vol. 39, pp. 252-5, May 2001.

[24] J. F. Veneman, R. Ekkelenkamp, R. Kruidhof, F. C. T. van der Helm, and H. van der Kooij, "A Series Elastic- and Bowden-Cable-Based Actuation System for Use as Torque Actuator in Exoskeleton-Type Robots," *The International Journal of Robotics Research,* vol. 25, pp. 261-281, Mar 2006.

[25] J. F. Veneman, R. Kruidhof, E. E. Hekman, R. Ekkelenkamp, E. H. Van Asseldonk, and H. van der Kooij, "Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation," *IEEE Trans Neural Syst Rehabil Eng,* vol. 15, pp. 379-86, Sep 2007.

[26] C. J. Walsh, D. Paluska, K. Pasch, W. Grand, A. Valiente, and H. Herr, "Development of a lightweight, underactuated exoskeleton for load-carrying augmentation," in *Proc IEEE Int Conf Robot Autom*, Orlando, FL, 2006, pp. 3485-3491.

[27] D. A. Winter, *The Biomechanics and Motor Control of Human Gait : Normal, Elderly and Pathological*, 2nd ed. Waterloo, Ont.: University of Waterloo Press, 1991.

[28] S. K. Banala, S. K. Agrawal, A. Fattah, V. Krishnamoorthy, H. Wei-Li, J. Scholz, and K. Rudolph, "Gravity-Balancing Leg Orthosis and Its Performance Evaluation," *IEEE Transactions on Robotics,* vol. 22, pp. 1228-1239, Dec 2006.

[29] M. Ishii, K. Yamamoto, and K. Hyodo, "Stand-Alone Wearable Power Assist Suit -Development and Availability-," *Journal of Robotics and Mechatronics,* vol. 17, pp. 575-583, Oct 2005.

[30] K. Yamamoto, K. Hyodo, M. Ishii, and T. Matsuo, "Development of Power Assisting Suit for Assisting Nurse Labor," *JSME International Journal Series C* vol. 45, pp. 703-711, 2002.