Rehabilitation Exoskeletal Robotics

The Promise of an Emerging Field

BY JOSÉ L. PONS

Exoskeletons are wearable robots exhibiting a close cognitive and physical interaction with the human user. These are rigid robotic exoskeletal structures that typically operate alongside human limbs. Scientific and technological work on exoskeletons began in the early 1960s but have only recently been applied to rehabilitation and functional substitution in patients suffering from motor disorders. Key topics for further development of exoskeletons in rehabilitation scenarios include the need for robust human–robot multimodal cognitive interaction, safe and dependable physical interaction, true wearability and portability, and user aspects such as acceptance and usability. This discussion provides an overview of these aspects and draws conclusions regarding potential future research directions in robotic exoskeletons.

The U.S. Department of Defense became interested in developing the concept of a powered suit of armor, whereas at Cornell Aeronautical Laboratories, the concept of man–amplifiers as manipulators to enhance the strength of a human operator was proposed. Efforts in the defense and military applications of exoskeletons have continued up to the present, mainly promoted by the U.S. Defense Advanced Research Projects Agency.

Rehabilitation and functional compensation (substitution) is another classic field of application for exoskeletons. Passive orthotic or prosthetic devices, dating back a few centuries, cannot be regarded as true robotic exoskeletons, but they may be regarded as the forebears of current rehabilitation exoskeletons [1].

Renewed interest in these old ideas is now leading to a well-established scientific and technological discipline attracting more and more research teams. However, can it be considered a mature research field? As discussed by Krebs et al. [2], the infancy of therapeutic robotics is easily demonstrated, as a MEDLINE search before 1990 will return no articles on therapeutic robotics. When looking at therapeutic, rehabilitation, or functional compensation exoskeletons, it becomes apparent that no articles on this matter are found in MEDLINE before 2000. Therefore, research on exoskeleton robots is just now beginning to fully emerge as an interesting field, particularly with regard to rehabilitation applications.

Because of the multipoint contact between both actors, the robot and the human user, interaction is expanding from a mere relay of information, e.g., as in teleoperation tasks or biofeedback, to a close physical and cognitive interaction. On the one hand, in rehabilitation and functional substitution exoskeletons, the physical human–robot interaction is to be understood as the generation of supplementary forces to empower and overcome human physical and motor limits [3],

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resulting from a disease or trauma. This involves a net flux of power between the exoskeleton and the human user.

On the other hand, the role of the cognitive human–robot interaction in rehabilitation exoskeletons is to make the human user aware of the possibilities of the robot (biofeedback) while allowing the user to relay intentions to control the robot at all times. Both physical and cognitive interactions have a direct impact on dependability. Take for instance the case of a lower-limb exoskeleton to compensate for weak muscles following postpolio syndrome. The user needs to relay his intention to start walking, and then the exoskeleton has to identify when assistance is required, e.g., for stabilization of the knee during stance. Risk of fall is not ethically acceptable; therefore, the cognitive interaction on which this relay of intentions is based must be 100% reliable. For example, it is not acceptable that the robot interprets that the limb is in swing when it is actually in stance. Following through with this example, if stabilization forces are applied through soft tissues, then it is also not ethically acceptable that these forces lead to damage of bodily structures, including skin, muscles, tendons, vessels, or nerves at the robot–limb interface. Therefore, the physical interaction is equally important in terms of dependability.

To provide a comprehensive analysis of the field, the "Relaying Information Between the User and the Exoskeleton" section addresses the cognitive interaction required to relay information between the two actors, the "Transfer of Power" section presents an analysis on the physical interaction, and the last section "Usability and Acceptance" focuses on an analysis of usability and acceptance as the true key requirements for dependability.

Relaying Information Between the User and the Exoskeleton

Relaying information on intention and volition (in terms of desired motor tasks) to drive, command, and control a robot is a difficult task whenever it has to be achieved in a natural way. Intention to move is a cognitive process that may comprise high-level functions carried out by the human brain. The functions include comprehension, visual perception and construction, the ability to calculate, attention (information processing), memory, executive functions (such as planning, self-monitoring, and perception), and motor control. In the context of motor control, this cognitive process leads to planning and execution of motor action, thus involving activity at central structures (i.e., brain and spinal cord) and peripheral structures (i.e., motor units and the musculoskeletal system), eventually leading to coordinated limb movement. Therefore, in principle, valuable information to decode human intention can be obtained from all these processes.

New trends are moving in the direction of sourcing the information directly from the human cognitive processes involved in the normal execution of tasks. These are called *natural interfaces*. The rationale for this approach comes from [4]:

- *Biological Reasons*: The interface systems seek to take advantage of the natural control mechanisms fully optimized in humans. Moreover, a lot of information is lost in the translation of biologically executed tasks into discrete events, e.g., natural movements or gestures into buttons or joysticks.
- *Practical Reasons*: Delays are introduced when natural cognitive processes are encoded into an imposed sequence of tasks. In addition, a training phase is needed to teach the user to generate these nonnatural commands or to map a cognitive process into a new set of outputs. Both delays and the mapping can induce fatigue in the user, both at a musculoskeletal and mental level.
- *Rehabilitation*: One of the main applications of exoskeletons, and in particular the one that we focus on in this article, is rehabilitation. Interacting directly with the phenomena involved in the cognitive process is a means to excite them and assess the evolution of the rehabilitation therapy.

In the following paragraphs, we will briefly highlight the most salient characteristics of systems for relaying commands to the exoskeletal robot while sourcing the information from the stages of motor planning to motor execution. In this regard, the "Brain-Controlled Exoskeletons" section focuses on implantable and noninvasive interfaces for sourcing the information from the central nervous system (CNS). The "Neural-Controlled Exoskeletons" section focuses on interfaces based on information obtained at the peripheral nervous system (PNS), and the "Movement-Controlled Exoskeletons" section addresses human–robot cognitive information from the analysis of human limb movements.

Brain-Controlled Exoskeletons

Research in brain–machine interface (BMI) technology has expanded dramatically in the last decade, with impressive demonstrations of nonhuman primates and humans controlling robots in real time through signals collected from cortical areas. These demonstrations can be divided largely into two categories: either continuous control of position or velocity [5], [6] or discrete control of more abstract information such as intended targets, intended actions, and onset of movements [7], [8].

There are several important questions regarding the control of artificial actuators directly from brain-derived signals. These include the type of brain signals (single unit, multiple unit, or field potentials) that would provide the optimal control signal for the device and the number of channels that may be necessary to operate a BMI efficiently for long periods of time. These and other questions were investigated in [5], where it was shown how macaque monkeys learned to use a BMI to reach and grasp virtual objects with a robot.

Single- and multiple-unit recording interfaces are only attainable through implantable microarray electrodes. These

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interfaces allow large data throughput but suffer from a number of shortcomings that limit their usability including:

1) the properties of the CNS are not prone to the design of effective but not too invasive interfaces
2) the implantation procedure is carried out blindly with no possibility of selecting the final position
3) biological reactions, e.g., gliosis, lead to nonstable interfaces
4) advanced algorithms need to be developed to extract useful neural information.

As a consequence, surface and noninvasive BMIs based on field potentials are much more common in practical control of rehabilitation exoskeletons. Research activities in this field are being strongly supported by the European Commission through its Information and Communication Technologies Programme (ICT), especially in the framework of the e-inclusion strategic objective. Several projects on BMI-controlled exoskeletons for rehabilitation and functional compensation have been launched, in particular, the BETTER project, "Brain-Neural Computer Interaction for Evaluation and Testing of Physical Therapies in Stroke Rehabilitation of Gait Disorders," on BMI-controlled lower-limb exoskeletons for rehabilitation of pathological gait in stroke survivors.

The principal goal of BETTER is to improve physical rehabilitation therapies of gait disorders in patients suffering from stroke, based on BMI-based assistive exoskeletons. The project goal is to validate (technically, functionally, and clinically) the concept of improving stroke rehabilitation with wearable exoskeletons and robotic gait trainers based on a top–down approach. The robot exerts physical stimulation (at the periphery) as a function of targeted neural activation patterns. This intervention is expected to result in reorganization within the cortex. Such top–down therapeutic treatment would aim to encourage plasticity of the affected brain structures to improve motor function.

One of the innovative aspects of the BETTER concept is the combined use of electroencephalogram (EEG) and electromyography (EMG) to relay information for controlling the lower-limb exoskeleton. EEG activity can be used in this way for discrete command of the exoskeleton. The combined use of EEG and EMG helps to detect involuntary movements, e.g., spasms, so that the exoskeleton can counteract them.

Therefore, even though implantable BMIs appear as an interesting alternative for relaying control information to exoskeletons, important drawbacks related to implantation and biostability of the implant lead to widespread use of surface technologies based on field potentials. These can only be used in discrete control of exoskeletons.

Neural-Controlled Exoskeletons

Natural control of wearable robots has been recently approached by directly interfacing with the human PNS. This is the case with the cyberhand system [9]. Several neural interfaces have been developed with different characteristics. Cuff electrodes have proven very reliable and robust, and they have the advantage of reduced invasiveness but suffer from limited selectivity. Sieve electrodes could present a very interesting solution for the development of neuroprosthetic and hybrid bionic systems, but still there are a number of problems limiting their usability [10], and they are only applicable to sectioned nerves.

A number of research groups have started investigating the possibility of developing and using neural interfaces featuring needles that are inserted longitudinally (LIFE electrodes [11]) or transversally (USEA electrodes [12], [13]) in the PNS. This approach looks very promising because it combines reduced invasiveness with good selectivity.

Even though intraneural PNS interfaces look very interesting and good results have been achieved during preliminary experiments in bidirectional control of hand prostheses [13], they suffer from similar drawbacks to implantable BMI interfaces (see the "Brain-Controlled Exoskeletons" section). Again, this leads to the use of surface EMG signals for practical control of exoskeletal structures.

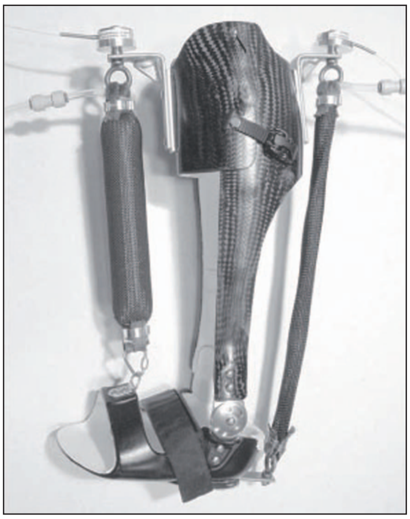

Fig. 1. Ankle–foot exoskeleton. Artificial muscle tension is controlled by EMG sensors in the natural muscles. (Adapted from [1].)

This is the case of the exoskeleton depicted in Figure 1. Proportional myoelectric control was implemented by means of a desktop computer and real-time control board. The controller regulated air pressure in the artificial pneumatic muscles proportionately to the processed muscle activation pattern. The artificial plantar flexor and tibialis anterior muscles were activated by soleus EMG and the artificial dorsiflexor muscle by tibialis EMG. EMG signals from the soleus and tibialis anterior were high-pass filtered with a second-order Butterworth filter (cutoff frequency of 20 Hz) to remove movement artifact as well as full-wave rectified and low-pass filtered with a second-order Butterworth filter (cutoff frequency of 10 Hz) to smooth the signal. Threshold cutoffs eliminated background noise, and adjustable gains scaled the control signals. When the low-pass filtered

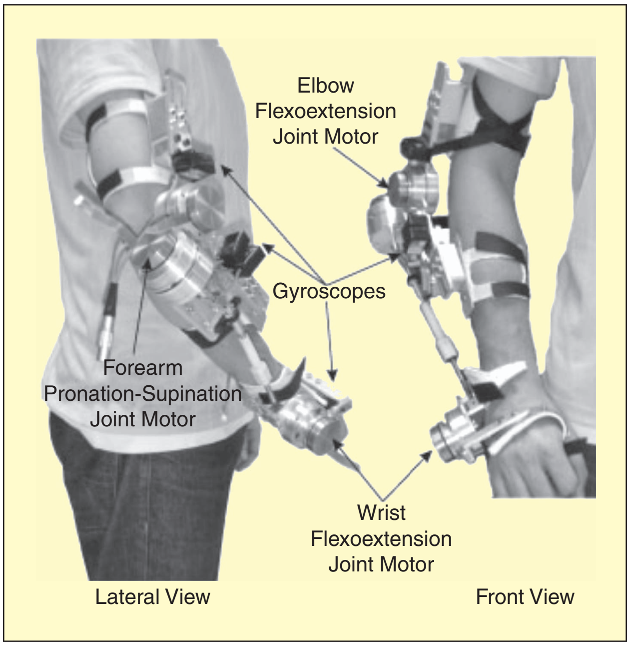

Fig. 2. An exoskeleton for controlling of three human upper-limb movements: flexion–extension of the elbow, flexion–extension of the wrist, and pronation–supination of the forearm in tremor supression applications. (Adapted from [1].)

soleus EMG signal was above threshold, the software inhibited all activation of the artificial dorsiflexor muscle.

Movement-Controlled Exoskeletons

The last step in the motor control process in humans is motor execution. Also, information obtained at this step can be used to relay control signals to the exoskeleton. This will be illustrated through the example of the upper-limb exoskeleton for tremor suppression depicted in Figure 2. Tremor is a rhythmic movement disorder resulting from some neurological conditions. When an exoskeleton is used to mechanically suppress tremor, ideally it should only apply loading to involuntary movements. Therefore, the cognitive interaction between the exoskeleton and the human wearer must determine whether the user is trembling or not, and if the user is trembling, the amplitude and frequency of the tremor must also be determined.

Certainly, this can be derived from EMG bursts driving tremor in the limbs, but it can also be determined from rhythmic movement of the upper limbs. In this regard, the controller of the exoskeleton in Figure 2 measures limb motion by means of inertial measurement units. This movement is fitted to a harmonic model that provides amplitude and frequency of tremor. Then, the exoskeleton applies a canceling force out of phase to the pathological tremor.

Similar to this example, motion sensors can be used in a number of instances to relay control commands to exoskeletons. The trend in combining command information from the three sources mentioned earlier meets the objective of increasing robustness in determining user intention.

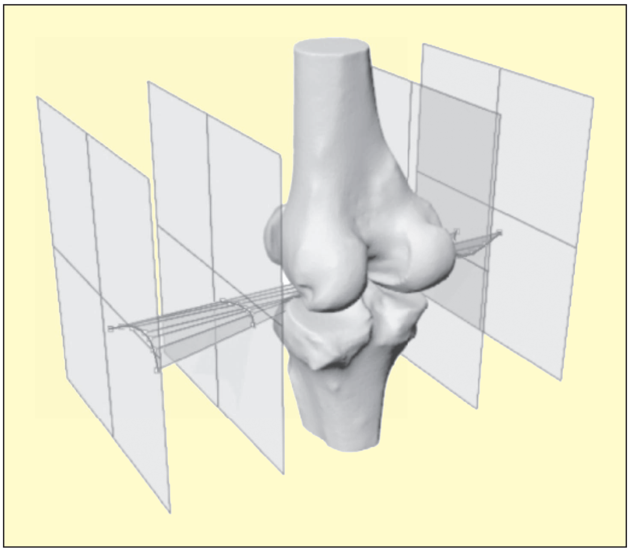

Fig. 3. Representation of three-dimensional movement of the ICR in a complete flexion–extension of the knee joint. (Adapted from [1].)

Transfer of Power

Motor rehabilitation and functional compensation (substitution) of motor disorders involves the transfer of power between the exoskeleton and the human user. This interaction, in its simplest manifestation, implies a physical coupling between the robot and the human, leading to the application of controlled forces between both actors. The actions of the two agents must be coordinated and adapted reciprocally since unexpected behavior of one of them during interaction can result in severe injuries. A classic example of physical interaction is exoskeleton-based functional compensation of human gait. Here, the robotic exoskeleton applies functional compensation by supporting human gait, e.g., by stabilizing the stance phase. This coordination implies kinematic compatibility between the exoskeletons and the human limbs' anatomy and the proper control of the interaction by means of advanced actuators.

Kinematic Compatibility Between Exoskeleton and Limb Anatomy

A good wearable exoskeleton design starts with the choice of a suitable kinematic structure for the device. That is to say, even before implementing actuation and control of the exoskeleton, the purely mechanical structure must enable wearability, ease of use, and operator comfort.

Once an exoskeleton has been implemented in hardware, there are many effects that can contribute to kinematic incompatibility between a wearable exoskeleton and a real human limb. The reason for that is the real-life variability of biomechanical parameters between subjects and also the variability of some parameters within individual subjects during movement. Unpredictability of joint axis locations and body segment sizes can disturb the interaction between an exoskeleton and the human operator, depending on the exoskeleton's kinematic design. This especially applies to exoskeletons that are wearable and kinematically equivalent to the human limb. To illustrate this, Figure 3 shows the projection of the knee instantaneous center of rotation axis (ICR) onto lateral planes (where the exoskeletal structure can be placed relative to the human limb) parallel to the human's sagittal plane. It is apparent that a revolute joint will not fit the ICR for all flexion angles. This will immediately lead to undesired interaction forces between the user and exoskeleton at the supports.

Natural control of wearable robots has been recently approached by directly interfacing with the human peripheral nervous system.

Typical biomechanical effects that cannot easily be captured within a human arm model used for exoskeleton development include:

- the intersubject variability of human limb parameters, e.g., D–H parameters such as length of bones, distances between rotation axes, and orientations of rotation axes
- the variability within an individual subject of joint centers of rotation during movement; the ICRs of each anatomical joint move very little during joint motion
- the intersubject variability of body segment dimensions: mass, size, and volume.

The impact of the kinematic configuration of the exoskeleton on the quality of the force interaction between the user and exoskeletons has been recently addressed in detail by Schiele and van der Helm [14]. It is concluded that compatible kinematics leading to a high-quality force interaction are best approached by introducing redundant joints in the exoskeleton rather than trying to exactly match human joints.

Actuator Selection and Design

In many instances, control strategies for exoskeletons require force-controlled actuators. An ideal force-controllable actuator would be a perfect force source, delivering exactly the commanded force independent of load movement. All force-controllable actuators have limitations that result in deviations from a perfect force source. These limitations include impedance, stiction, and bandwidth.

Dynamic performance of actuators is important in any exoskeleton application, but it is critical in functional compensation (substitution) for the disabled. Within the latter two application domains, lower-limb exoskeletons, such as those used to assist gait, are the ones that impose the strictest requirements in terms of power and torque delivery.

In exoskeletal robotics, traditional actuator technologies, e.g., pneumatic, hydraulic, and electromagnetic actuators, are commonly used. Hydraulic and pneumatic actuators are known for their high force density and high force or torque characteristics and have been used in a number of applications. Figure 4 illustrates a soft-actuated full-body exoskeleton for use as a force extender [15]. The system is based on pneumatic muscle actuators and benefits from the compliant actuation characteristics of pneumatic actuators. This compliant behavior is basically due to the high compressibility of the actuator fluid.

Direct drive actuators are also a good approximation of an ideal force source. However, they are too large for exoskeletons that must support the wearer's weight in addition to the weight of the actuators. Therefore, their use is limited to applications where the actuator can be placed in a nonmoving base of the exoskeleton, thus strongly constraining wearability and portability. Transmission stages are then required.

Other authors [16] use conventional actuator technologies, e.g., electromagnetic dc motors, to build series elastic actuators. This is done in the context of a variable impedance exoskeleton for functional compensation of gait in wearers affected by drop foot. The actuator is used to drive the exoskeleton in the sagittal plane and is set to control the applied force and the compression in the spring through the motor's angular rotation. The resulting actuator exhibits low impedance, low friction, and an acceptable dynamic range.

In other instances, when unrestricted wearability and portability is a requirement, passive versions of exoskeletons are the solution. This is the case of the GAIT exoskeleton [17]. In this particular exoskeleton (Figure 5), the system has been developed for functional compensation (substitution) in patients with postpolio syndrome where muscle weakness at the knee level is found. In this particular case, the role of the knee actuator includes stabilizing the knee during the stance phase of the gait cycle and assisting the knee extension at the end of the stance phase so that the knee is fully extended at heel contact.

When modeling the torque versus knee flexion angle, a linear relationship is found during stance that basically corresponds to a linear spring. Under these specific circumstances, the GAIT exoskeleton implements linear rigid springs to substitute natural muscles during stabilization of stance and linear soft springs for assisting knee extension. This requires virtually no energy input (even though energy is still needed to switch between both springs), and portability is met.

Finally, new actuation alternatives are emerging as functional electrical stimulation (FES) systems that are gaining better performance. In rehabilitation and functional compensation

(a) (b)

Fig. 4. Power-assist suit incorporating pneumatic actuators and exhibiting superb compliance characteristics. (Adapted from [1].)

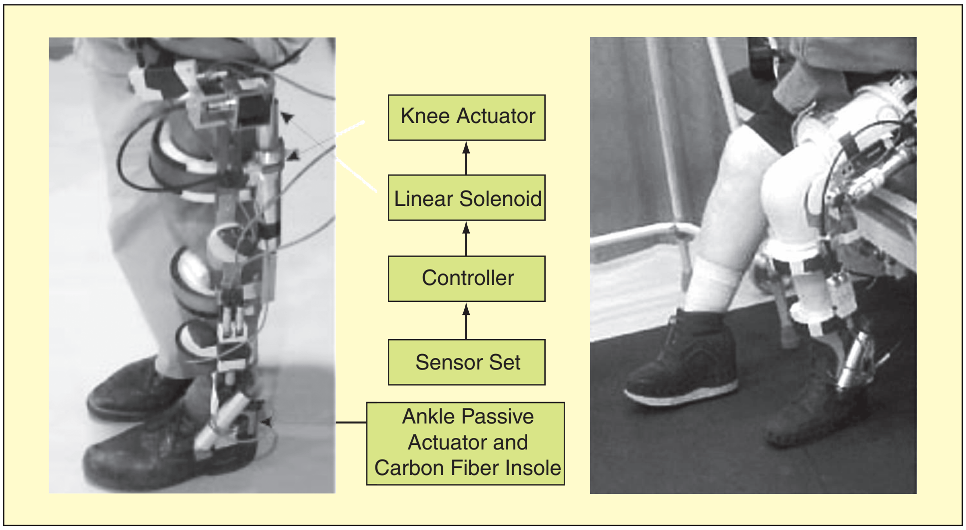

Fig. 5. Passive portable exoskeleton: springs are used to model the mechanical behavior of the musculoskeletal system at knee and ankle level. (Adapted from (1).)

(substitution) applications, it is common that musculoskeletal systems and structures are preserved after a lesion, leading to motor impairments. This is what happens in the case of stroke, where a brain lesion may affect motor areas in the brain, but in principle, peripheral nerves and muscles are still undamaged or are in pathological tremor where central oscillators drive the limbs in involuntary rhythmic movements, but muscles are preserved.

Under these circumstances, it is possible to consider using natural muscles as actuators, provided that enough selectivity is achieved and also provided that intelligent use of the FES system is attained so that fatigue and pain limits are not reached. Figure 6 shows the FES version of the exoskeletal robot in Figure 2. It is apparent that, by introducing FES, very compact solutions can be attained with the positive impact on wearability, portability, and, as a consequence, acceptance and usability.

Usability and Acceptance

Excessive pressure is one of the main concerns related to the application of loads to the body. The application of loads by the exoskeleton to the musculoskeletal system produces contact pressures that can compromise safety and comfort. These two aspects are closely related to acceptance and usability. In this regard, two aspects relating to the pressure applied have been defined: pressure distribution and pressure magnitude. The former relates to comfort, whereas the latter relates more to safety. Regarding safety, the usual guideline is to avoid pressure above the ischemic level, i.e., the level at which the capillary vessels are unable to conduct blood, thus compromising the tissue. This level has been estimated at 30 mmHg [18].

The relationship between applied pressure and comfort is complex. Some authors use a combination of peak pressure, pressure gradients, and contact area to quantify discomfort [19]. Touch receptors are sensitive to deformation of the layers of tissue where they are located [20], and therefore, the perception of pressure is indirect: pressure deforms tissues, and this deformation is sensed by skin receptors. Furthermore, the type, density, and distribution of skin receptors vary significantly from one part of the body to another. Finally, the skin receptors respond dynamically to excitation (receptor adaptation). This dynamic response means that pressure perception is dependent on the dynamics of the process whereby the pressure is applied.

The design of exoskeletons implies the design of structures intended to apply loads to the human skeleton through the layers of soft tissue between the supports of the robotic system and the bones of the user. There are two main aspects that must be taken into account when designing support systems for wearable robots, including identifying what are the anatomical areas and structures able to support effective loads and the maximum levels of pressure that these structures can handle without raising issues of safety and comfort. This immediately leads to the definition of pressure tolerance areas, which, in turn, result in maps where high-, middle-, and low-tolerance areas are represented both for the lower and upper limbs onto which the exoskeleton is applied (see Figure 7 for the upper limb).

Conclusions

Despite several decades of work on exoskeletons, significant scientific contributions in the application domain of rehabilitation and functional compensation and substitution began to appear only in the last ten years.

As exoskeletons are characterized by close cognitive and physical interaction with the human user, requirements put to the cognitive interaction are strict. In most instances, only 100% accurate decoding of user intention is a valid figure when relaying command information to the robot. It is likely that future research activities will seek robust cognitive interfaces by sourcing control information from various stages of the neuromotor control structure of the wearer.

Also, physical interaction is of importance to ensure dependability. As exoskeletons apply actions by exerting

Fig. 6. Textile-based version of a tremor suppression exoskeleton based on FES systems to replace artificial actuators.

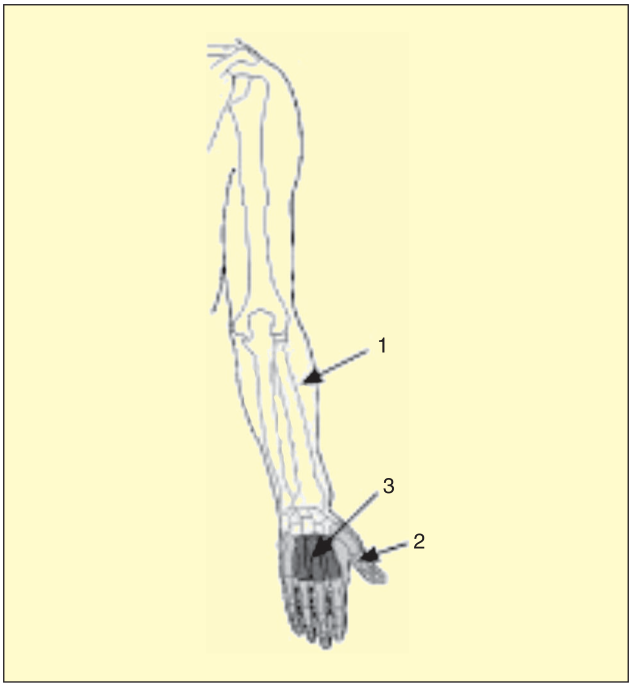

Fig. 7. Areas of sensitivity to pressure in the hand and forearm. 1: low-tolerance area (average, approximately 450 kPa), 2: middle-tolerance area, and 3: high-tolerance area (average, approximately 950 kPa).

forces through soft tissues, kinematic compatibility and quality of the physical interaction need to be addressed. If, in addition, this is to be implemented in fully portable embodiments, actuators become key technologies. Research will probably evolve toward hybrid technologies, combining exoskeletons and motor neuroprostheses. In this regard, FES will appear as a promising alternative to artificial actuators.

Finally, exoskeletons will be widely accepted only for rehabilitation and functional compensation if the user is a central agent in the design, development, and validation of these technologies. Only under these conditions will exoskeletons be accepted and usability proven. To this end, cosmetics, aesthetics, and dependability are key factors to be considered. In this particular regard, the field of rehabilitation exoskeletons is not mature enough.

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