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UPPER LIMB POWERED EXOSKELETON

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An exoskeleton is a wearable robot with joints and links corresponding to those of the human body. With applications in rehabilitation medicine, virtual reality simulation, and teleoperation, exoskeletons offer benefits for both disabled and healthy populations. Analytical and experimental approaches were used to develop, integrate, and study a powered exoskeleton for the upper limb and its application as an assistive device. The kinematic and dynamic dataset of the upper limb during daily living activities was one among several factors guiding the development of an anthropomorphic, seven degree-of-freedom, powered arm exoskeleton. Additional design inputs include anatomical and physiological considerations, workspace analyses, and upper limb joint ranges of motion. Proximal placement of motors and distal placement of cable-pulley reductions were incorporated into the design, leading to low inertia, high-stiffness links, and back-drivable transmissions with zero backlash. The design enables full glenohumeral, elbow, and wrist joint functionality. Establishing the human-machine interface at the neural level was facilitated by the development of a Hill-based muscle model (myoprocessor) that enables intuitive interaction between the operator and the wearable robot. Potential applications of the exoskeleton as a wearable robot include (i) an assistive (orthotic) device for human power amplifications, (ii) a therapeutic and diagnostics device for physiotherapy, (iii) a haptic device in virtual reality simulation, and (iv) a master device for teleoperation.

Keywords: Activities of daily living; dynamics; human arm; exoskeleton; kinematics; wearable robotics.

1. Introduction

Integrating capabilities of humans and robotic-machines into a unified system offers numerous opportunities for developing a new generation of assistive technology. Potential applications could benefit members of both healthy and disabled populations. For many physical tasks, human performance is limited by muscle strength. Similarly, muscle weakness is the primary cause of disability for persons with a variety of neuromuscular diseases including stroke, spinal cord injury, muscular dystrophies, and other neuro-degenerative disorders. Opposite this limitation in muscular strength, humans possess specialized and complex algorithms for control of movement, involving both higher and lower neural centers. These algorithms enable humans to perform very complicated tasks such as locomotion and arm movement, while at the same time avoid object collisions. In contrast, robotic manipulators can be designed to perform tasks requiring large forces or moments, depending on their structure and on the power of their actuators. However, the control algorithms that govern their dynamics lack the flexibility to perform in a wide range of conditions while preserving the same quality of performance as humans. It seems therefore that combining these two entities, the human and the robot, into one integrated system under the control of the human, may lead to a solution which will benefit from the advantages of each subsystem. The mechanical power of the machine, integrated with the inherent human control system, could allow efficient performance of tasks requiring higher forces than the human could otherwise produce. At the heart of this human-machine integration lie two fundamental scientific and technological issues: (i) the exoskeleton (orthotic device) mechanism itself and its biomechanical integration with the human body, and (ii) the human machine interface (HMI). These two key issues will determine the quality of the integration between the human and the exoskeleton in terms of how natural it will be for the operator to control the exoskeleton device as a biological extension of his/her body.

The exoskeleton, as an assistive device, is an external structural mechanism with joints and links corresponding to those of the human body. The human wears the exoskeleton, and its actuators generate torques applied on the human joints. In utilizing the exoskeleton as a human power amplifier, the human provides control signals for the exoskeleton, while the exoskeleton actuators provide most of the power necessary for task performance. The human becomes part of the system and applies a scaled-down force in comparison with the load carried by the exoskeleton. Using the exoskeleton as a master device in a teleoperation system enables the operator to control a secondary, possibly remote, robotic arm (slave). In a bilateral mode, the forces applied on the remote robotic arm by the environment are reflected back to the master and applied on the operator's arm by the exoskeleton structure and actuators. In this setup, the operator feels the interaction of the robotic arm tool-tip with the environment.[1,2] Employing the exoskeleton as a haptic device is a relatively new technology aiming to generate human interaction with virtual objects simulated in virtual reality. As a result, virtual objects can be touched by the operator. The exoskeleton structure and its actuators provide force feedback, emulating the real object including its mechanical and texture properties. The exoskeleton, in that sense, simulates an external environment and adds the sense of touch (haptics) to the graphical virtual environment.[1,3–5] Several mechanisms including arms,[6–18] hands,[19,20] legs,[21–24] and other haptic devices were developed for many applications.[25]

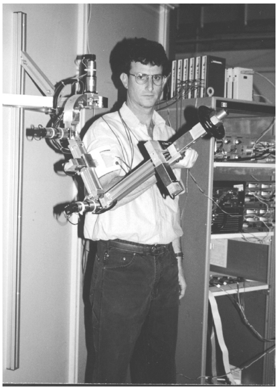
Throughout the last three decades, several designs of exoskeletons for human power amplification have been developed and evaluated. In studying the evolution of these systems, two basic types with different Human Machine Interfaces

(HMI) seem to emerge, which may be defined as generations. The first exoskeleton generation was developed based on the mission profile of the US Department of Defense that defined the exoskeleton as a powered suit to augment the lifting and carrying capabilities of soldiers. It was originally named "man-amplifier". The primary intent was to develop a system that would greatly increase the strength of a human operator while maintaining human control of the manipulator. The first generation prototype, known as Hardiman, was the first attempt to mechanically design a man-amplifying exoskeleton using a hydraulically powered articulating frame worn by an operator.[26–30] The position-controlled architecture led to poor responsiveness and instability. The second generation of exoskeletons placed the HMI at the dynamics level, utilizing the direct contact forces (measured by force sensors) between the human and the machine as the main command signals to the exoskeleton. The human wore the extender, in a way that linked them together mechanically. The operator was in full physical contact with the exoskeleton throughout its manipulation.[22–24,31–34] Several experimental extender prototypes were designed and built in order to study issues associated with this mode of control.

A common feature in both the first and second generation of exoskeletons was that the operator must apply an action, either kinematic — position command (first generation) — or dynamic — contact force command (second generation) — in order to trigger the exoskeleton response. Obviously, this sequence of events constitutes a source of delay in both systems. The inherent time delay of the physiological system when combined with the delay of the system's mechanical actuators has a deleterious effect on overall performance. From the control theory perspective, gain and time delays are linked together. Inherent time delays in a system reduce the phase margin, and hence stability will be reduced in addition to inherently limited bandwidth.[35] Moreover, from the operational perspective, the operator feels as if the exoskeleton is following his/her limb movements rather than moving with his/her limb synergistically. As a result of a limited operational bandwidth in these first generations, the operator will not be able to act and respond quickly, for example, to catch a falling object.

Setting the HMI at higher levels of the human physiological (neurological) system hierarchy, one can overcome the electro-chemical-mechanical delay, usually referred to as the electro-mechanical delay (EMD).[36] This inherent time delay refers to the interval between the time when the neural system activates the muscular system and the time when the muscles and the associated soft tissues mechanically contract and generate moments around the joints. The main advantage of establishing the interface at the neuromuscular level is the ability to estimate the forces that will be generated by the muscles, using a muscle model, before the muscle contraction actually occurs. As a result, the reaction time of the human/machine system should decrease, resulting in a more natural control of the task. In line with this concept, a third generation of exoskeletons may be defined by setting the HMI at the human neuromuscular junction.

(a)



(b)

(c)

Fig. 1. Three generations of upper limb exoskeletons: (a) 1-DOF exoskeleton, elbow joint, (b) 3-DOF exoskeleton — two degrees of freedom at the shoulder joint and one DOF at the elbow joint — and (c) the CADEN-7 — two exoskeleton arms (left and right arm) with seven degrees of freedom each.

One of the principle innovative ideas of the research was setting the HMI at the neuromuscular level and utilizing processed surface electromyographic (sEMG) signals as one of the primary command signals of the system (Fig. 1). The exoskeleton as an assistive device will act as a "*human-amplifier*" increasing the force generated by the human muscles. As a human amplifier the exoskeleton will provide the operator a scaled-down version of the external load. For example, in case of an object manipulation, the human may feel only 10% of the external load while the exoskeleton carries the remaining 90% of the load. Although this general concept may be applied to other parts of the human musculoskeletal system, the upper limb (arm) was selected for the purpose of studying the hypotheses associated with the proposed HMI of the exoskeleton. The upper limb is composed of segments linked by articulations with multiple degrees of freedom and is able to perform tasks requiring both power and precision of movement. From the technological perspective, an electromechanical system that is fully portable is still not feasible, due to the power to weight ratio of the power source. However, for the human upper limb, portability is not as critical of an issue as it is for the lower limb. The exoskeleton system may be part of a stationary working station, or it could be fixed to the frame of a powered wheelchair and powered by the wheelchair battery.

The primary hypothesis of a neural-controlled exoskeleton is that the HMI can be set at the neuromuscular level, using processed sEMG signals as the primary control signal to the exoskeleton. These signals are the same signals initiated by the human's central nervous system to contract the human's own actuators (the muscles). The primary component that enables this approach is the *myoprocessor.* The myoprocessor is a set of computational representations (models) of the human arm muscles running in real-time and in parallel to the physiological muscles that predicts joint torque.

The scope of this manuscript is to provide an overview of a research effort to develop, integrate, and study a seven degree-of-freedom (DOF) exoskeleton and to explore its operation as a human amplifier utilizing a neural controller.

2. Methods

Two sets of databases were collected prior to the design of the exoskeleton arm. The kinematics and the dynamics of the human arm during activities of daily living (ADL) were studied in part to define the engineering specifications for the exoskeleton arm design. In addition, the neural activities in well-structured single and multi-joint arm activities, under various loading conditions, were studied in order to develop muscle models (myoprocessor) that further enable development of the neural-control approach of the exoskeleton.

2.1. *Human arm kinematics — Experimental protocol*

The kinematics and the dynamics provide the fundamental criteria for designing a seven DOF exoskeleton that is aimed at supporting the operational workspace and functionality of the human arm. The kinematics of the human arm were collected during 24 ADL tasks from six subjects ranging in age from 20 to 41 years. Mean and standard deviation of height, weight, and age for the subjects was 1.72 ± 0.08 m, 76.2 ± 23.1 kg, and 26.2 ± 7.7 years, respectively. Of the six subjects, three subjects were male and three were female. The arm movements included four sub-groups: (i) three-dimensional reach-pick-place objects, (ii) eating and drinking, (iii) hand–body interactions associated with hygiene, (iv) functional activities including an interaction with a door, phone, pitcher and a cup: Arm kinematic data was collected using a VICON motion capture system (Vicon Inc.) at a sampling frequency of 120 Hz. Raw optical data was captured synchronously from 12 cameras and filtered with a Woltring quintic spline filter having a mean square error of $20\,\mathrm{mm}^2$. Reflective markers of 14 mm diameter were attached to the right (dominant) arm of each

(a) (b) (c)

Fig. 2. Human arm kinematics — Experimental setup: (a) Reflective markers attached to the upper arm; (b) subject sitting at the table setup performing a reach-pick-place action; (c) human arm joints, links, and coordinate assignment for the Vicon system (Human model from Bodyworks — Zetec, Ltd. Inc).

subject at seven key anatomical locations (Fig. 2). Individual models and marker sets were calibrated for each subject and further used during Vicon post-processing of subject data.

A seven DOF model of the human arm was developed including three segments (upper arm, lower arm, and hand) connected to each other and the human trunk with a frictionless ball-and-socket shoulder joint, two-axis elbow, and two-axis wrist. Using this model, seven equations of motion were derived. The mass, the center of mass location, and the inertia of the human arm segments were estimated for each subject. The general form of the equations of motion is expressed in Eq. (1).

$$\tau = M(\Theta)\ddot{\Theta} + V(\Theta, \dot{\Theta}) + G(\Theta), \tag{1}$$

where $M(\Theta)$ is the 7×7 mass matrix, $V(\Theta, \dot{\Theta})$ is a 7×1 vector of centrifugal and Coriolis terms, $G(\Theta)$ is a 7×1 vector of gravity terms, and τ is a 7×1 vector of the net torques applied at the joints. Given the kinematics of the human arm $(\ddot{\Theta}, \dot{\Theta}, \Theta)$ the individual contributions on the net joint toque (τ) vector were calculated for each action of each subject. These allow one to examine the contribution of the individual components (inertial, centrifugal, and Coriolis, and gravity) to the net torque.

2.2. *Exoskeleton arm — Design methodology*

Two exoskeleton arms were developed, each including seven degrees of freedom (DOFs). Each exoskeleton arm is actuated by seven DC brushed motors (Maxon) that transmit the appropriate torque to each joint utilizing a cable-based transmission system. Four force/torque sensors (ATI Industrial Automation) are located at all interface elements between the human arm and exoskeleton as well as between the exoskeleton and external load, measuring all forces and torques acting and reacting between the human arm, the external load, and the exoskeleton [Fig. 1(c)]. The mechanisms are attached to a frame mounted on the wall, which allows both height and distance between the arms to be adjusted.

Articulation of the exoskeleton is achieved about seven single-axis revolute joints: One for each shoulder abduction-adduction (abd-add), shoulder flexion-extension (flx-ext), shoulder internal-external (int-ext) rotation, elbow flx-ext, forearm pronation-supination (pron-sup), wrist flx-ext, and wrist radial-ulnar (rad-uln) deviation. The exoskeletal joints are labeled 1 through 7 from proximal to distal in the order shown in Fig. 3(a). Note that the order is consistent with the axes presented in Fig. 2(c).

The fundamental principle in designing the exoskeleton joints is to align the rotational axis of the exoskeleton with the anatomical rotation axes. If more the

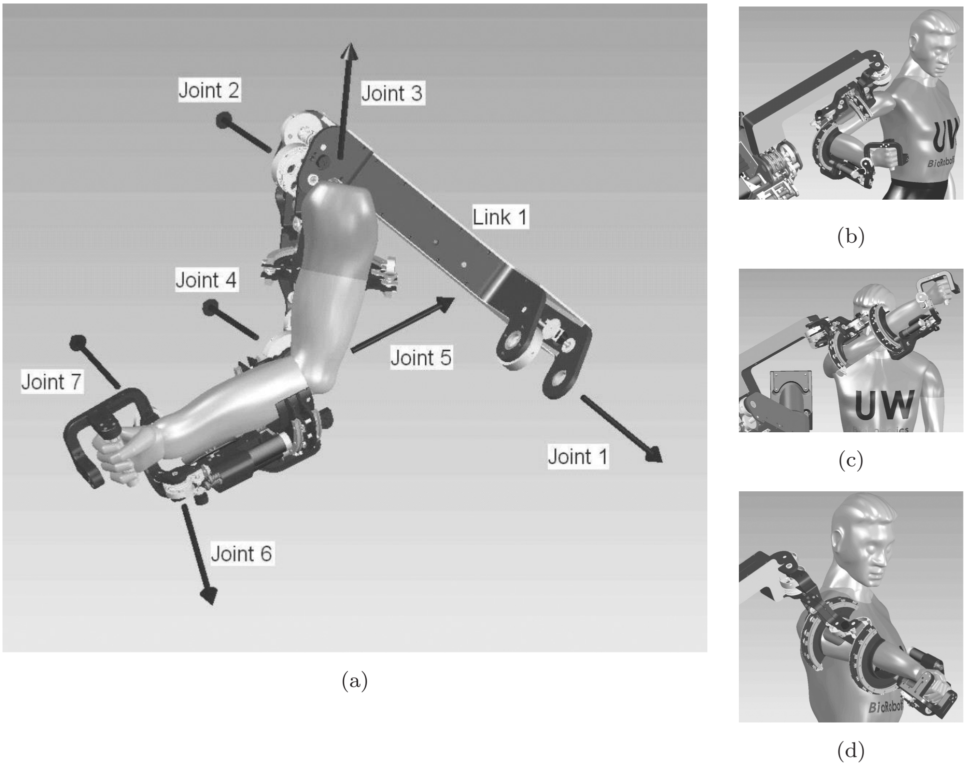


Fig. 3. The exoskeleton joint definitions and the mechanism singular configurations: (a) CAD model (Solidworks, Concord) of exoskeletal axes assignment in relation to the human arm. Singular configurations between axes 1 and 3 occur around the shoulder int-ext rotation axis in configurations (b) and (c). A singular configuration between axes 3 and 5 occurs in full elbow extension (d).

one axis is involved at a particular anatomical joint (e.g. the shoulder and wrist), the exoskeleton joints emulating the anatomical joint intersect at the center of the anatomical joint. Consistent with other work, the glenohumeral (G-H) joint is modeled as a spherical joint composed of three intersecting axes.[37] The elbow is modeled by a single axis orthogonal to the third shoulder axis, with a joint stop preventing hyperextension. Exoskeletal prono-supination takes place between the elbow and wrist joints as it does in the physiological mechanism. And finally, two intersecting orthogonal axes represent the wrist. Comparing the range of motion of the anatomical joints in the ADL study and the range of motion of the exoskeleton shows that the exoskeleton's joints support 99% of the ranges of motion required to perform daily activities.

Upper arm and forearm rotations present special challenges in design as a result of the human arm occupying a large volume along the joint axis of rotation. This space, which in classical robotics would be occupied by mechanical joints, must be kept not only clear of obstacles but also of component configurations that could result in injury or discomfort to the user, such as sharp edges or pinch points. Semi-circular bearings, achieving a laterally-open interface, were used to allow users to don the device without strain or discomfort. On the adjacent link distal to these semi-circular axes, the mechanical components of the HMI are mounted. The mechanical HMI (mHMI), as opposed to the neural HMI, consists of a pressure-distributive structural pad that is rigidly mounted to a six-axis force/torque sensor and simultaneously securely strapped to the mid-distal portion of each respective arm segment.

The open design of the mHMI is a beneficial feature in assistive applications with mobility-impaired users, but the attachment of the interface itself is most critical. The assumption that the human arm can be represented as a seven DOF system is an assumption pertaining to skeletal and end-effector biomechanics. Components such as skin, musculature, and other soft tissues cannot be represented by a model as simple as a similar seven DOF representation. This is a significant observation given that most exoskeleton research, including the research presented in this text, uses non-invasive techniques for the mHMI. All non-invasive interfaces will have imperfections that are influenced by the selection of interfacial size, position, and orientation. mHMIs that are placed excessively high (proximal) on the arm, may produce unnecessarily large forces, and potential discomfort to the user, through the interface points when operated in an assistive mode. Additionally, compliance of relaxed musculature in proximal regions of the limb, as well as non-uniform transformations during muscular contraction, reduce interfacial stiffness and produce higher non-linear disturbances in force measurement, which would ultimately result in reduced bandwidth of performance. Cross-sections of distal parts of limb segments are less variable in magnitude and experience fewer underlying skeletal transformation, making them better apt for mHMI attachment.

Another significant consideration in exoskeleton design is placement of singularities. A singularity is a device configuration where a DOF is lost or compromised as a result of alignment of two rotational axes. The fundamental principle in addressing the singularity of the exoskeleton is, if possible, to place singular configurations, through joint design, outside or at the edge of the anthropometric reachable workspace of the human arm. For the exoskeleton arm, singularities occur when joints 1 and 3 or joints 3 and 5 align [Figs. 3(b) and 3(c)]. The singularity between joints 3 and 5 naturally occurs only in full elbow extension, i.e. on the edge of the forearm workspace [Fig. 3(d)]. Each of these singular configurations take place at or near the edge of the human workspace leaving the majority of the workspace free of singularities. Moreover, for optimal ease of movement in any direction, singular axes should be placed orthogonal to directions where isotropy is of highest importance. For the singularity placement shown in Fig. 3, isotropy will be maximized at 42.5° of shoulder flexion and 26.4° of shoulder abduction, values that lie in the median of shoulder ROM based on the ADL study.

In the field of wearable robotics, weight is a critical factor that frequently must be sacrificed for the sake of strength or rigidity. However, development of a rigid structure that lacks adequate bandwidth is as ineffective of a tool as one that is lightweight but lacks structural rigidity. To achieve both rigidity and bandwidth, critical decisions were made regarding transmission type and placement of actuators.

Placing the motors, the heaviest components in the exoskeleton system, has a significant impact on the overall performance of the system and the quality of its interaction with the human operator. Motors for joints 1–4 were mounted on the stationary base, achieving a 60% reduction in overall weight of the moving parts. The remaining three motors, whose torque requirements are substantially less, were positioned on the forearm. As each motor carries the weight and inertia of the more distally placed motors, the importance of high power-to-weight ratio increases from shoulder to wrist. Shoulder and elbow joints are each driven by a high torque, low power-to-weight motor (6.23 Nm, 2.2 Nm/kg), while wrist joints are driven by a lower torque, high power-to-weight motor (1.0 Nm, 4.2 Nm/kg). The primary advantage of using cable drive transmissions lies in their ability to transmit loads over long distances without the friction or backlash inherent to gears. The absence of backlash is achieved through the structural continuity of the cable, enabling a direct link between the driving shaft and the shaft or link being driven. Cable-driven systems, including one stage of speed reduction for joints 5–7 and two stages of speed reduction for joints 1–4, were developed and integrated to transmit torques from the actuators to the various joints.[38] An I-beam cross section shape was used for all the links allowing bilateral cable routing, as well as high structural stiffness and appropriate strength.

2.3. *The myoprocessor and neural control*

Two basic conditions have to be fulfilled in order to establish an HMI at the neuromuscular junction. The first condition is the capability to measure the bio-signals (myosignal intentions of muscles contraction) involved in the joint's movement, which can be measured using surface electrodes, a non-invasive technique. The second condition is the ability to simulate and to predict the functions of the human body's subsystems and organs from the interface level (myosignals) down to the lower levels of the physiological hierarchy (skeletal muscle forces and moments). The term "myoprocessor" is used to define the component of the system that simulates the human skeletal muscles behavior and provides an estimation of the muscle forces.[39] The myoprocessor runs in real-time and in parallel to the physiological muscles.

During the electro-chemical-mechanical time delay (EMD), the system gathers information regarding the physiological muscle's neural activation level based on processed EMG signals, the joint positions, and joint angular velocities. This information is fed into a myoprocessor, which in turn predicts the moments that are going to be developed by the physiological muscles relative to each joint. The predicted moments are fed to the exoskeleton system such that by the time the physiological muscle contracts, the exoskeleton has amplified the joint moment by a pre-selected gain factor and assisted the movement. Part of the time gained by using these predicted muscle forces is employed by the electromechanical subsystems of the powered exoskeleton to compensate for their own inherent reaction time. Figure 4 depicts the fundamental building block of setting the HMI at the neural level. The main advantage of establishing the interface at the neuromuscular level is the ability to estimate the forces that will be generated by the muscles before the muscle contraction actually occurs. This concept takes advantage of the inherent delay (EMD) of the musculoskeletal system. The EMD refers to the interval of time between neural activation of the muscular system and the generation of moments around the joint from the mechanical contraction of musculature and associated soft tissue [Fig. 4(b)].

Hill based muscle models (myoprocessor) were developed as the core component of the exoskeleton control algorithm. The performance evaluation of the myoprocessor was based on a database that experimentally collected input/output signals of the myoprocessor. The experimental setup included passive weight lifting machines that were used for single and multiple upper arm joint movements. The data collection included the joint kinematics and dynamics in addition to sEMG signals collected simultaneously from 28 muscles at 1 KHz (Fig. 5) under various loading conditions (25%, 50%, and 75% of the maximal isometric loading condition). Two types of models were developed and analyzed: (i) Hill based models[39,40] and (ii) Neural Networks.[40] Genetic algorithms were further used to tune the internal parameters of the Hill-based model that predicted the joint torque within a variance of 15% of the nominal value.

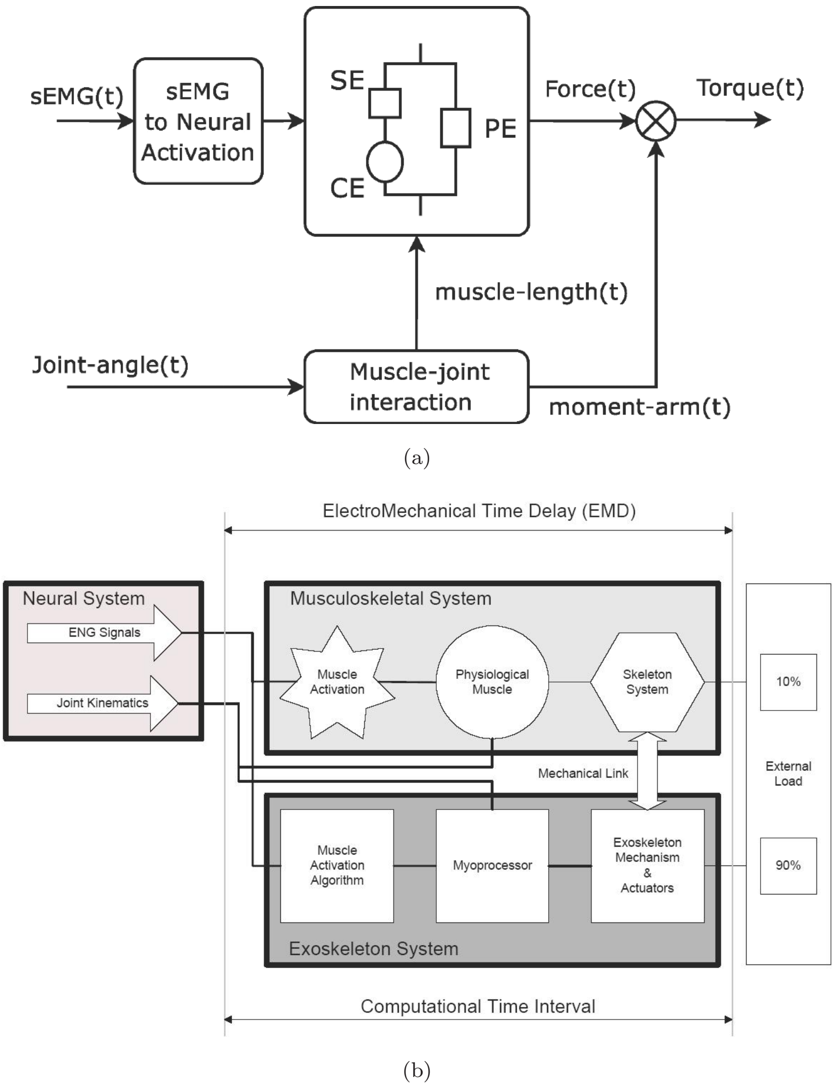


Fig. 4. The myoprocessor: (a) low-level block diagram of a hill-based muscle model, and (b) system-level block diagram of neural control scheme for the exoskeleton system.

2.4. *Exoskeleton performance evaluation protocol*

The exoskeleton performance was previously conducted in static and dynamic conditions with a fixed external load.[41] The current experimental protocol focused on the exoskeleton performance evaluation under dynamic loading conditions that were unpredictable from the operator's perspective. These loading conditions were simulated by generating sudden changes in the external load. The subject was wearing the exoskeleton in a standing posture with his arm fully stretched and perpendicular

Fig. 5. Experimental setup for measuring neural activity (sEMG) along with the kinematics and dynamics of selected arm activities (e.g. flexion and extension of the elbow) under various loading conditions.

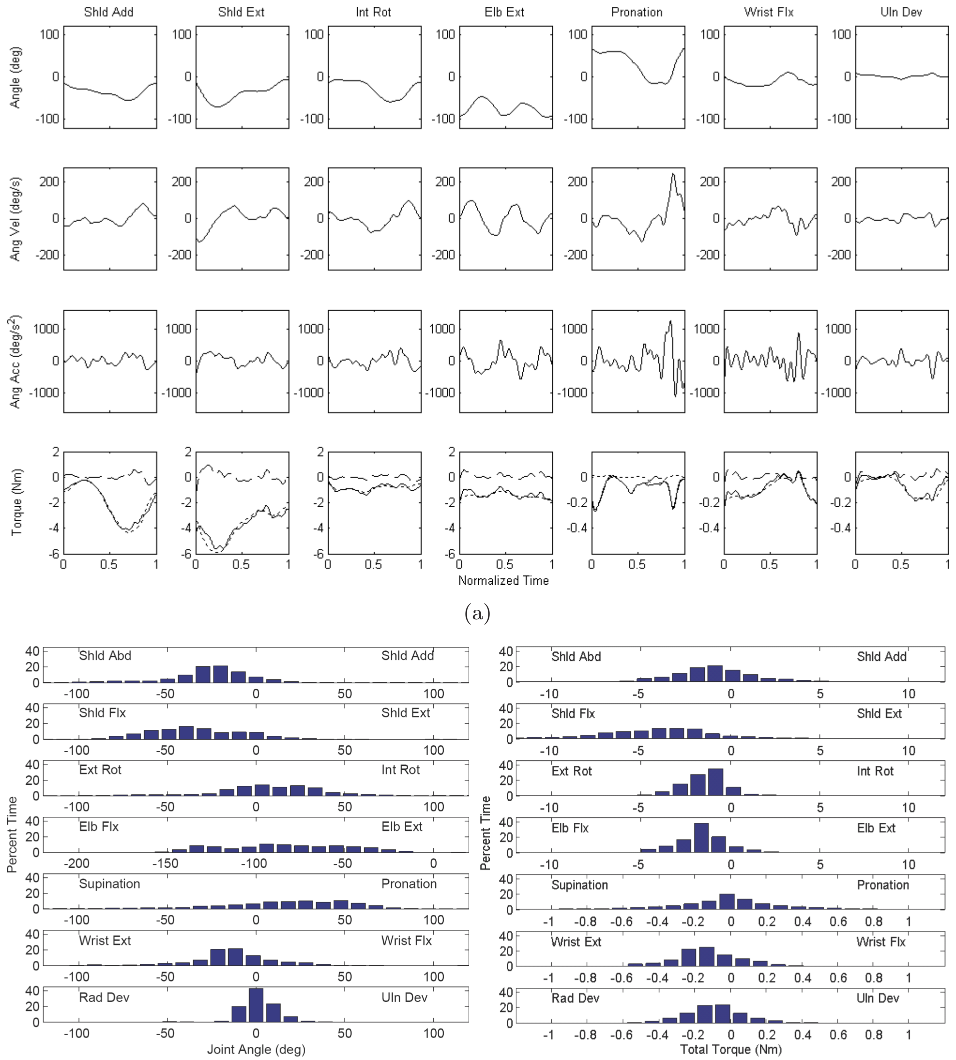
to the ground. The exoskeleton operator was instructed to generate a full elbow flexion movement followed by a full elbow extension while caring an external load of 1 kg. When the elbow angle reached 90°, an additional external mass of 1.265 kg was added to the exoskeleton. As a result, in the elbow joint angular range of 0–90°, the external exoskeleton load was 1 kg and in the angular range of 90–150° the external load was 2.265 kg. The mass was added during the flexion movement and released during the extension movement at the elbow joint angle of 90°.

3. Results

3.1. *Human arm kinematics*

The statistical distribution of the human arm kinematics, collected experimentally during ADL's, along with the dynamics calculated based on the analytical model are summarized in Fig. 6. The kinematics and the dynamics of the arm while performing a functional arm reach task are plotted in Fig. 6(a). The last row of Fig. 6(a) depicts, for each joint, the total joint axis torque as well as the gravitational terms and the inertial, centrifugal, and Coriolis terms combined. In general, the magnitude of joint torque is small for the distal joint and higher for the proximal joints. Regardless of the joint location, the joint torque component that compensates the gravitational loads is significantly larger than the inertial, centrifugal, and Coriolis terms combined.

In fact, the overall shape of the joint axis torque during the selected daily activities is dictated solely by the low frequency gravitational term. This phenomenon was observed in all the actions included in the ADL database that were collected in



(b) (c)

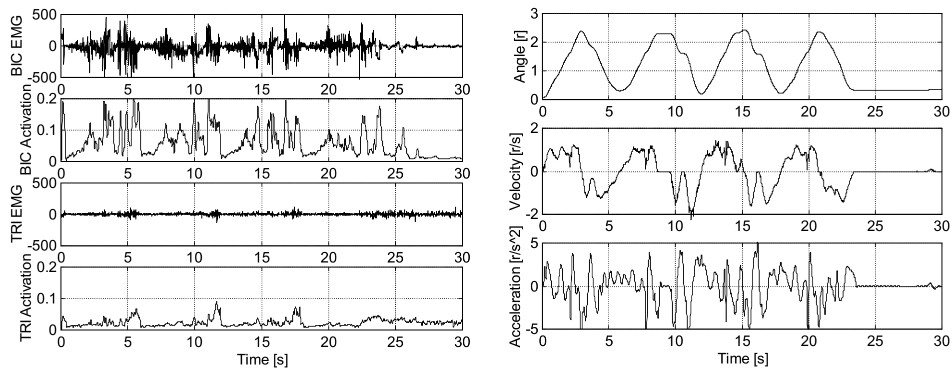
Fig. 6. Kinematic and dynamic data of the human arm during ADL's: (a) Time histories of the joint angles and the net joint torques during a functional arm reach motion (opening a cupboard door), where joint torques (τ) are separated into total torque (solid), torque due to gravity (dotted), and torque due to inertial, Coriolis and centrifugal effects (dashed); statistical distribution of (b) joint angles and (c) joint torques are displayed as histograms, plotted sequentially from the top as Vicon axes 1 through 7. Zero position of the arm shown in Fig. 2(c).

this study. Identifying the joint torque component due to gravitational loads as the largest component provides the justification for incorporating a gravity compensation algorithm not just for the exoskeleton alone but for the human operator arm as well as part of the control algorithm of the system.

While some distributions appear normal in shape, others possess a bi-modal or even tri-modal form where modal centers correspond to key anthropomorphic configurations [Figs. 6(b) and 6(c)]. These configurations are positions of the arm that occur commonly throughout daily activities, often where joint velocities are low at the initial or final periods of motion trajectories.

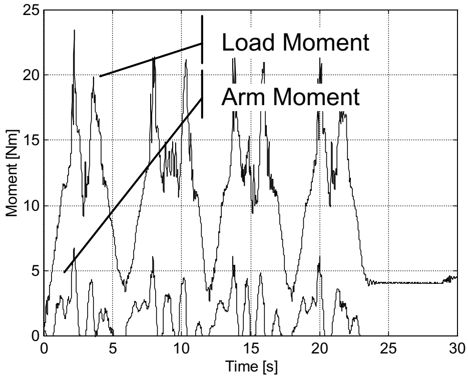
3.2. *Exoskeleton performance*

The data collected experimentally under dynamic and unpredictable loading conditions indicate a natural and stable movement of the integrated human/exoskeleton system (Fig. 7). The EMG signals and the activation levels of the Biceps Brachii (BIC) and the Triceps Brachii (TRI) muscles [Fig. 7(a)] show a high level of activity during the two transient periods of increasing and decreasing the load. The level of muscle activity is higher during the process of decreasing the load in both the BIC and the TRI muscles. Those high muscle activation levels are accompanied with velocity decrease during the elbow flexion movement, aimed to preserve stable smooth movement in inertial transient processes [Fig. 7(b)]. The dynamic loading transient process appears as a spike phenomenon in both the human arm and the load moments [Fig. 7(c)] indicating the sudden increase/decrease of the load during the flexion/extension movement, respectively. The exoskeleton gain is constant during the elbow flexion/extension movement; however, it tends to change in a bounded range during transient loading phenomena [Fig. 7(d)]. The straight line in Fig. 7(d) represents a constant mechanical gain. The 45° inclination of the line represents a mechanical gain of one (non-assistive mode of the exoskeleton).

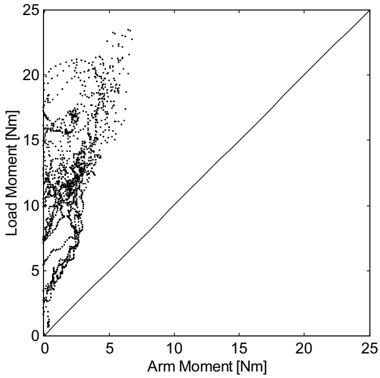


(a) (b)

Fig. 7. Exoskeleton operational performance in unpredicted, dynamic loading conditions: (a) raw EMG signals and muscle activation levels of the Biceps Brachii (top) and the Triceps Brachii (bottom); (b) elbow joint kinematics; (c) arm and external load moments about the elbow joint, where the difference between the two moments is generated by the exoskeleton; (d) external load moment as a function of the human arm moment.



(c)



(d)

Fig. 7. (*Continued*)

4. Conclusions and Discussion

Developing a powered exoskeleton that would serve a human operator adequately during daily activities requires a profound understanding of the kinematics and dynamics of the human arm during these activities, and is beyond the anthropometric information that has been widely known for several decades.[42,43] Expanding existing knowledge, the kinematics and dynamics of the human arm were studied in order to provide the engineering specification to facilitate the design of a seven DOF exoskeleton arm. The results indicate that the various joints' kinematics and dynamics change significantly based on the nature of the task. Gross position is usually achieved by utilizing the shoulder and the elbow joints with relatively low joint velocities. However, fine manipulations are performed by the wrist and forearm rotations with higher joint velocities. As expected, the joint torques decrease as we move from the proximal end of the arm to the distal end, which is also correlated with the muscular mass of the arm.

Analyzing the contribution of individual terms of the arm's equations of motion indicate that the low frequency gravitational term is the most dominant term in these equations. The magnitudes of this term across the joints and the various actions is higher than the inertial, centrifugal, and Coriolis terms combined. In particular, the effects of velocity-dependent terms are commonly two orders of magnitude smaller than those dependent on position (gravity). Coincidentally, the velocity-dependent terms are computationally the most expensive terms in the equations of motion, contributing more than half of the total execution time. These computationally expensive terms can legitimately be eliminated from the dynamic compensation in the control algorithm. Eliminating these terms significantly reduces execution times of command signals without negatively impacting system performance. The results justify using only the gravitational terms of the equations of motion as a compensator in part of the control system.

The seven DOF exoskeleton design that followed the study of the kinematics and the dynamics of the arm in ADL's relied heavily on its findings to provide the required engineering specifications. Additionally, principles of physiological joints assisted in achieving a relatively lightweight, high-performance system facilitating full-workspace and ROM, as defined by the ADL study. Proximal placement of motors, distal placement of pulley reductions, and open mechanical human-machine-interfaces were incorporated into the design of the exoskeleton. Additional characteristics include low inertias, high-stiffness links, and back-drivable transmissions without backlash.

The HMI is one of the key factors in creating an exoskeleton that is perceived and controlled by the operator as a natural extension of his/her own body. It is a point of the integrated human machine system in which energy and information are exchanged between the two subsystems. The HMI may include two types of interfaces that can, in part, merge into a single bioport: (i) the mechanical port which serves as the physical interface between the human body and the exoskeleton system, and (ii) the information port which generates the command signals for the exoskeleton by the operator. Following biomechanic and anthropometric principles led to an exoskeleton design that is compatible with the human anatomy. The mechanical design therefore satisfies the requirements for the mechanical port. Setting the information port at the neural level aims to use the same control signals that are used by the human body for controlling the exoskeleton subsystems. The optimal location of the information port at the neural hierarchy is still an open scientific question. A low-level bioport, such as invasive or non-invasive electrodes collecting neural or myosignals (EMG), utilizes a muscle-modeling approach and takes advantage of the inherent electromechanical delay to predict the physiological muscle response. A high-level bioport, such as invasive neural signals collected directly from the motor cortex or a non-invasive approach utilizing EEG signals, relies on the plasticity of the motor cortex to learn and adjust its command signals while avoiding the profound complexity involved in understanding the function of the neural system. There is no doubt that as research progresses in the biocompatibility of invasive neural electrodes, coupled with a deeper understanding of the neural system, more viable bioport will emerge for a superior HMI.

From the electromechanical perspective, both relatively high torque-to-weight ratio motors as well as the energy density of the exoskeleton energy source are the major limiting factors in developing portable exoskeleton system. It is recognized that in the case of a lower limb exoskeleton, the weight of the power source, as well as the actuators and structure, is transmitted to the ground without loading the human body, and therefore may overcome, in part, this technological deficiency. However, the functionality of an upper limb exoskeleton must be significantly compromised if the aim is to make the system completely portable.

The proposed myoprocessor enables a neural interface between the human operator and the exoskeleton system. This neural interface contributes to a natural and stable integration between the wearable robot and its operator such that the operator views the exoskeleton as an intuitive extension of his/her body. Further exploration of the myoprocessor and automatic adjustment of its internal parameters based on patient data will continue to be an active field of research. Using the exoskeleton as a human assistive device remains one among many potential applications of the system, including automatic physiotherapy, utilization as a haptic device for interaction with virtual objects in a virtual environment, or as a master device for teleoperation with force-feedback capabilities.

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