2011 IEEE International Conference on Robotics and Automation
Shanghai International Conference Center
May 9-13, 2011, Shanghai, China

Exoskeletal Spine and Shoulder Girdle for Full Body Exoskeletons with Human Versatility

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Abstract—**Currently, wearers of full body exoskeletons are hindered in their ability to use their upper body as desired due to the rigid back parts used in these devices. In order to maximize their versatility the design and preliminary testing is shown of an exoskeletal spine mechanism, called "exo-spine", that allows the wearer to move all degrees of freedom of his spine and shoulder girdle. Based on the primary forces to be supported during lifting, identified as gravity forces from loads lifted in front of the wearer, as well as functional degrees of freedom, which is a control strategy used by our central nervous system, this mechanism can be actuated using only one motor to provide the required support. Experiments indicate a substantial, although not problematic amount of friction as well as further requirements for the control of the assisting force. Besides improving exoskeletons its basic structure and design principles may be successfully applied to rehabilitation as well.**

I. Introduction

VERSATILITY, the ability to move freely in all directions, in particular of our upper body, is indispensible for the kind of work that people do [1]. For exoskeletons to become successful in assisting human activities, they will need to enable their wearers to solve the problems they face with the degree of versatility that they would normally have. If not, even basic tasks as lifting items from the floor will require unnatural body postures, become undoable and/or require a much greater amount of effort.

Currently, the part of full body exoskeletons between the hips and shoulders is completely rigid. This restriction on both the wearer's spine (flexion, lateral flexion, and rotation) and, to a lesser extend, his shoulder girdle (abduction and elevation) thus leave the wearer with limited versatility. Moreover, since alltogether these parts contain 7 degrees of freedom (DOF), exoskeletons would require 7 extra actuators, using standard robotics technology, to regain this movability. This paper therefore presents a novel mechanical solution called "exo-spine" that, by maximizing the effectiveness of its actuation to the achievement of heavy work and lifting assistance, enables the required augmentation using only one motor.

This introduction will continue to set the specific context based on which the exo-spine is both required and possible. Subsequent sections will explore the mechanics, control method, as well as experiments to investigate the friction and controlability, and finally the discussion.

A. Spinal and Shoulder Movability in Exoskeletons

Given the arrangement of DOF on the human body full versatility in exoskeletons is especially difficult to achieve in the upper body. Full arm actuation has been done, such as [2], although not yet in untethered, fully wearable types. As for the spine and shoulder, several devices exist that assist (parts of) the upper body; they can be grouped as follows.

Exoskeletons with an unlimited power supply include both wearable types with a tether as well as those fixed to a base [2-4]. Although wearability is restricted to the power supply, this group has fewer limitations on the amount of actuators. Two solutions for shoulder motion can be seen: free shoulders and arms with interaction at the hands [3], and full actuation using one motor per DOF [2] [4]. Another group consists of full body exoskeletons that carry their own power supply [5] [6]. With this extra limitation on the amount of actuators neither spine motion nor shoulder girdle motion has been implemented. More lightweight exoskeletal devices that attach to the arm are used for rehabilitation and force feedback systems [7] [8]. Their applications allow for a separate power supply and the required actuator forces are lower, such that full shoulder actuation is possible.

Comparing the above devices it can be concluded that the available power and actuators impose strong limitations on a battery powered exoskeleton. Moreover, the conventional solution of one actuator per DOF would require more motors than can be carried along.

B. The HAL Robot Suit

The current HAL (Hybrid Assistive Limb) suit, HAL-5, is a full body exoskeleton that carries its own power supply. It consists of frames interconnected by power units that each contain an electromotor and reduction gears and are positioned directly next to the hip, knee, shoulder (flexion) and elbow joints of the wearer to assist his movements [5]. Additional passive DoF are located at each shoulder, upper arm, and ankle joint. The suit is powered by batteries.

The system is controlled according to the intentions of the wearer, which are obtained by measuring the bioelectric signal (BES) on the skin above the main flexor and extensor muscles associated with each augmented human joint. Motor torques are calculated according to these signals. It is expected that similar control techniques and actuators will be used in versions that will contain the exo-spine.

This work was supported in part by the Grant-in-Aid for the Global COE Program on "Cybernics: fusion of human, machine, and information systems" at the University of Tsukuba and relevant organizations.

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C. Heavy Work and Lifting

Rosen *et al.* found that when performing daily living tasks the mean joint torques were at least an order of magnitude larger than those torques without the gravitational component [9]. In addition, HAL, like many augmenting exoskeletons, is meant to assist lifting during heavy work tasks, such that gravity forces will account for almost all required actuation. Furthermore, the weights of objects that are likely to be lifted with an exoskeleton are too high, and the sizes too large for the objects to be carried on one side while still being able to walk in a stable and balanced way.

Apart from gravity forces pulling and pushing forces may be found as well, such as seen in hospitals [10]. However, muscles at the main body joints from ankle to shoulder would all exert force in the same direction during both pulling and lifting, whereas pushing would be the opposite of lifting. Therefore, the kind of assistive forces the spine and shoulder girdle need most of the time are those that assist these parts to counter gravity forces from loads in the front. Instead of having as many actuators as the amount of DOF used for lifting it could thus be more effective to use a few actuators that focus on such lifting action only.

D. Functional Degrees of Freedom

To further combine multiple DOF into one it is possible to exploit a strategy, used by our central nervous system to control our high-DOF bodies in 3D space, called “functional degrees of freedom” (fDOF) [11]. An fDOF implies that in certain situations two or more muscles act based on the same control signal. For lifting, when first of all considering the static balances at the hip and the spine, and assuming there are no external moments on the lower back, it can be seen that the gravitational moments around the frontal axis at the hip and spine must be correlated. Furthermore, Thomas *et al.* have shown that during reaching tasks 94.7% of the peak-to-peak dynamic torques (i.e. excluding the gravitational components) at the ankle, knee, hip, spine, shoulder and elbow are determined by one parameter [12]. When considering only the hip, spine and shoulder the correlation will be even higher. Taken together it may be concluded that there is one fDOF that controls nearly all muscle torque of the hip and spine during lifting. Since HAL already uses hip motors that are controlled based on the BES of the hip muscles of the wearer, these signals can therefore be combined and used as a control signal for the exo-spine.

E. Applications for Spinal Cord Injury Patients

Spinal cord injury (SCI) patients have limited or no abilities to control the muscles that balance their pelvis, which even during sitting makes their upper body unstable [13]. They often solve this by leaning on an armrest or their knee with one arm, but doing so leaves them with only one hand to do most of their daily living tasks. Not only can the BES of the hips be used for the exo-spine, but similarly the BES of the upper back muscles of SCI patients may be effectively used to control a hip motor during sitting. Looking further ahead into this research, using the exo-spine and a hip motor as a hip and back support for SCI patients could increase the area they can reach with both hands when sitting in a wheelchair [14], as well as help them stand upright when such a back support would be attached to a lower body exoskeleton that assists their walking.

II. Design

This section will first describe the general design principles used followed by the design of a prototype based on these principles.

A. General Design

As mentioned, HAL uses BES based torque control of the motors to provide assistance; position control is completely left to the wearer, who can voluntarily change his BES to change both his own muscle torques and those of HAL’s motors. The same principle will apply to the exo-spine. Only the assistive torque needs to be determined for the wearer to be able to move the exo-spine as desired. Furthermore, HAL’s hip component is fastened to the wearer’s pelvis, so that as long as the top of the exo-spine is attached to and can follow the wearer’s shoulder girdle the shape of the exo-spine itself does not matter.

A further requirement follows from the position of the exo-spine behind the human spine. The exo-spine will have to become longer as it bends forward to ensure HAL’s shoulders remain lined up with the wearer’s shoulders.

As for HAL’s shoulder girdle movement there are two simplifications that can be made. They are based on a principle first applied by Schiele and Van der Helm [7] where two passive joints are inserted, perpendicular to the active joint, between the actuator and the attachment with the wearer, such that any misalignment between, e.g. the shoulder’s active joint and the wearer’s shoulder does not create painful forces between HAL and the wearer. Using such a system HAL‘s passive shoulder joint (arm medial rotation) may be placed behind the wearer’s shoulder instead of above. This opens up the space above the wearer’s shoulder so that he can freely elevate his shoulders when needed, accommodated by the added passive joints between the actuator and the attachment with the wearer.

Shoulder abduction can be provided by the exo-spine by adding exoskeletal “shoulder blades” that make the same forward rotation as the wearer’s collar bones, but behind the wearer, such that HAL’s shoulders can move forward with respect to the top of the exo-spine.

To save energy and, in particular, to counter the effects of friction, which will be further explained from Section III.B, springs can be added that balance with the weight of HAL’s upper body. This will relieve some of the actuator’s required power during both up and down bending.

B. Prototype

In order for the exo-spine to bend forward into a convex shape, similar to that of the wearer’s spine, as well as to extend simultaneously its basic mechanism follows the one shown in Fig. 1. Two types of parts, called “vertebra” and "link" are connected into a chain of in total five vertebras, including the base. The spine extends when bending forward as the instantaneous centers of rotation (ICOR) of all vertebras start in front of the exo-spine. Extension becomes slower as bending increases since the ICOR move closer to the vertebras (Fig. 1b). Despite the many parts the whole exo-spine bends forward as one single DOF.

Fig. 2 shows a drawing of the central vertebras (top) and links (bottom). (As in Fig. 1b numbers refer to the different vertebras; letters indicate the joints; X points to the front.) They are interconnected with rotational joints at A and rod ends, which provide 3 DOF motion, at joints B and C. This, and a parallelogram structure at each link that enables joint C to move sideways, allows the vertebras to rotate around their vertical axis. In addition, each vertebra can bend sideways (laterally) around the axis connecting B and C.

The full exo-spine is shown in Fig. 3: full bending and shoulder blades abducted (a), full side bending (b) and full rotation (c) (Height when straight is 350mm). When combined with HAL the base will be located just behind the 2nd lumbar vertebra of the wearer, L2, while the shoulder blades extend up to the top of his shoulder. The exo-spine can not hyperextend; this is blocked mechanically. Its neutral position is straight, without any lateral bending or rotation, similar to the human spine.

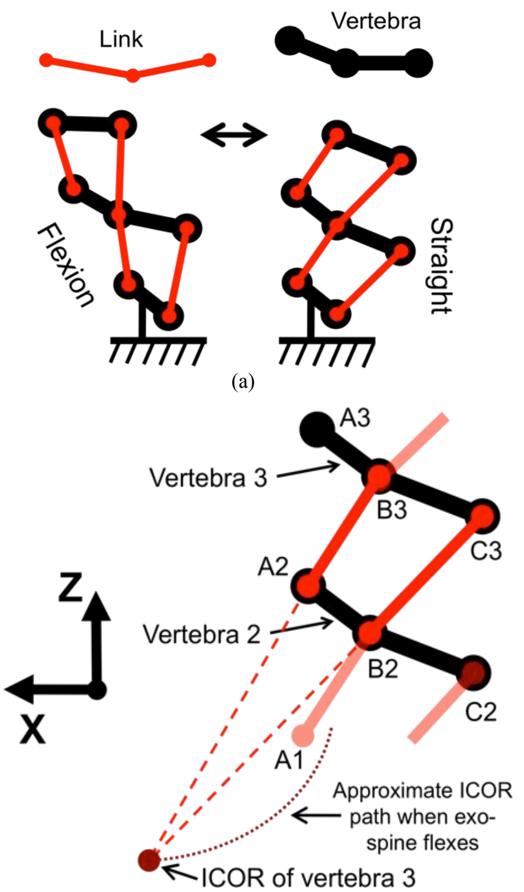


Fig. 1. Schematic side view of the vertebra-link mechanism of the exo-spine. As the mechanism bends each next vertebra rotates forward with respect to the one below (a). Moreover, the exo-spine as a whole extends when bending due to the fact that the instantaneous center of rotation (ICOR) of each vertebra starts in front of the spine.

III. Actuation

This section will describe the actuation method and primary control algorithm. Although verification of this algorithm is not included it will show how fDOF can be implemented into exoskeletons

A. Mechanism

The exo-spine is actuated using two cables that run over small pulleys in the back corners of the structure from the base to the top (Fig. 2). The cables are made of high strength Dyneema and pulling them generates a moment on the exo-spine that pulls towards the neutral position.

The cables are connected below the spine onto one pulley. When the exo-spine is bent laterally to one side the distance between the top and the base for the cable on the other side becomes larger such that only that cable actuates the exo-spine while the other becomes slack. This produces a torque that pulls back towards the neutral position. When the exo-spine rotates the pulleys of the vertebras and links move away from each other horizontally, so that the cables come into a zigzag shape. The tension on the cables then produces a torque that again pulls towards the neutral position. Based on the fDOF between the spine and shoulder during lifting, the two cables connect to a small lever at the top that in turn pulls the exo-spine's shoulder blades towards the zero abduction position. When lifting, assumed that it is in front

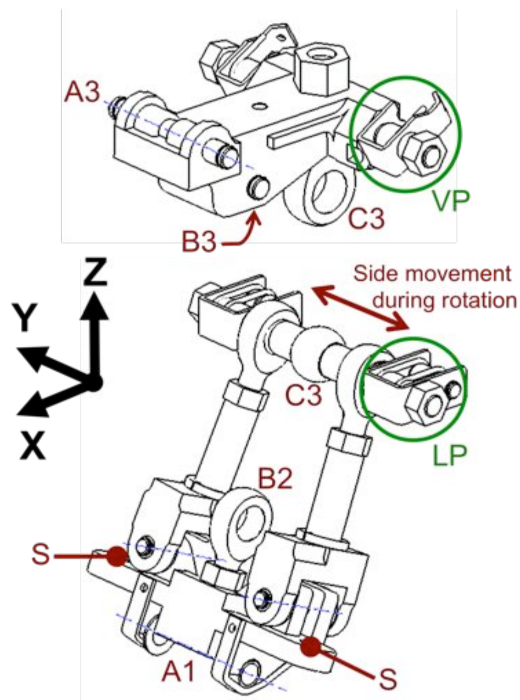
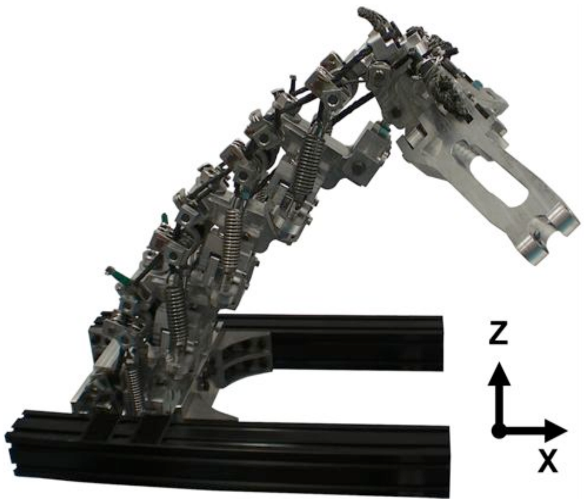
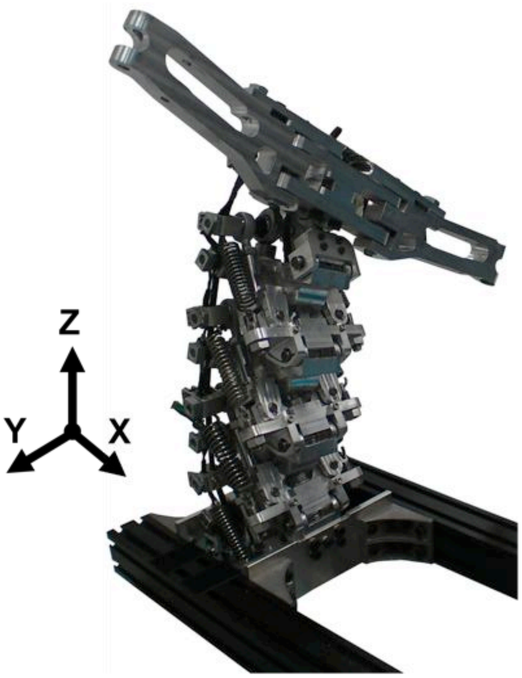


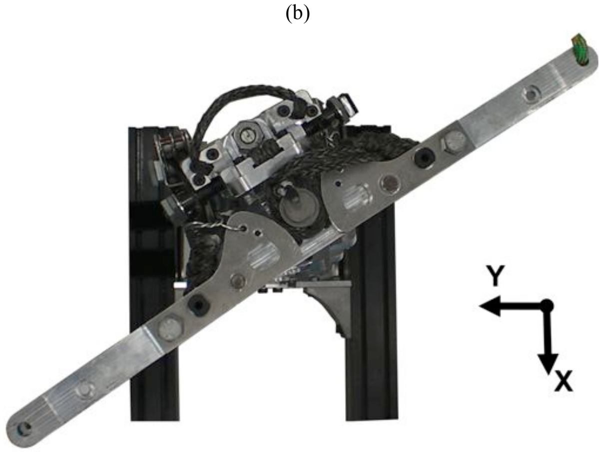
Fig. 2. CAD drawing of the actual vertebras (top) and links (bottom). Joints are indicated as: front (A), middle (B) and rear (C); the number indicates the vertebra to which they belong. These indications are as in Fig. 1b. The lower attachments of the springs are indicated by S, one vertebra pulley by VP, and one double link pulley by LP. All pulleys are located right above each other.



(a)



(b)



(c)

Fig. 3. Exo-spine with attached base (black) for testing. It is shown in full forward bending and shoulder girdle abduction (a), side bending (b) and rotation (c), but any combination of these is possible. Coordinate frames correspond with those of Fig. 1 and 2.

of the wearer, irrespective of the amount of bending, side bending and rotation of the exo-spine or abduction of the shoulder blades, the assistive torques will be counteracting forces that result from pulling as well as the gravity forces of the load on HAL's arms and pull toward the neutral position. Pushing however can only be done in the straight position as the cables can not produce any forward bending forces.

As mentioned above, only control of the top is important, the position of each vertebra is not controlled. This, however, can also lead to buckling, and thus increased friction. Although each vertebra has only a small range of motion (ROM) and buckling is limited to about 10mm deflection of the center, without countermeasures the exo-spine would always be buckled. Springs are therefore attached at the sides to both counter buckling (calculated based on the maximum load and deflection) as well as balance the weight of HAL laterally. This way, buckling is still possible, but will happen only occasionally. The springs are attached at "S" in Fig. 2 and connect to a short cable that crosses pulleys "LP" and is fixed next to "VP".

As for user safety, this can be ensured by blocking the cables at a certain length. For extra safety a backup cable runs through the center of the exo-spine.

B. Control

Forces on the cables are generated using a motor located below the base vertebra. Using the hip-spine fDOF as a basis for the control, the torque control signals of the hip motors added together, M_{hip}, become the control input signal for the exo-spine. Although likely possible, using the back muscle (erector spinae) BES would only add to HAL's setup time.

During usage the cable length is known (from the pulley angle), but the positions of the carried loads are not. Moreover, there will be a certain friction that reduces the required cable force, F_{cable}, when the exo-spine bends down, and increase the force when bending up. In addition, measuring the generated moment on the exo-spine, M_{spine}, is not possible due to friction as well as unknown forces from interactions with the wearer's body. However, assuming that

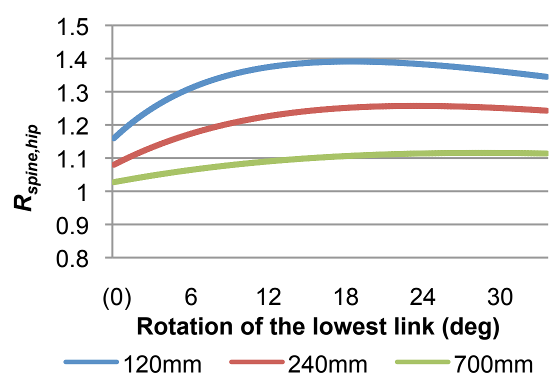


Fig. 4. Simulation results showing the ratio $R_{spine,hip}$ of the moment at the exo-spine to the moment at the hip versus the amount of bending of the exo-spine as measured by the rotation of the lowest link. This is shown for loads carried at different horizontal distances in front of the HAL's shoulder motor with the base of the exo-spine straight, and the shoulder girdle not abducted. As the controller can not measure the center of gravity of the load a certain distance must be assumed.

the friction is a linear function of M_{spine} and F_{cable}, it will be possible to do experiments to obtain two formulas, one for bending up and one for down, that give the ratio $R_{cable,spine}$, which is F_{cable} / M_{spine}, for a certain length of the cable below the base, L_{cable}.

To further calculate the required torque in the exo-spine, the ratio $R_{spine,hip}$, which is M_{spine} / M_{hip}, must be known. It is determined by the position of the center of gravity of the carried objects and HAL's arms with respect to the locations of the exo-spine's ICOR, which move as the wearer flexes (Fig 1b), and the center of rotation (COR) of the hip. That the center of gravity influences $R_{spine,hip}$ can be seen from the results of a SolidWorks Motion simulation comparing different horizontal distances from the load to the shoulder, as shown in Fig. 4. The further the load is held in front of the shoulders the lower $R_{spine,hip}$. A distance that is most likely found during usage must therefore be chosen.

With M_{hip} as the input signal the force F_{cable} becomes

$$F_{cable} = M_{spine} R_{spine,hip} R_{cable,spine}, \tag{1}$$

in which both ratios depend on L_{cable}, while $R_{cable,spine}$ also depends on whether the exo-spine bends up or down. It is possible to obtain this bending direction directly from the wearer's behavior. In addition, to save energy, when the exo-spine does not move the motor control should be in the "bending down" state, which gives a lower F_{cable} and combined with the friction will still be able to hold the load. When the wearer increases his hip BES for some short time while the exo-spine does not bend down he can be assumed to intend to bend up. As soon as there is no motion for some short time the wearer can be assumed to intend to hold still or bend down. A further compensation may be included to change $R_{spine,hip}$ according to the absolute angle of HAL's hip component, which is measured standard in each HAL suit.

Since feed-forward control is used and friction might change due to temperature or wear there will in addition be a "friction dial" that can be set by the user to let the controller assume a lower or higher friction. This will also be convenient for the user to set his own most comfortable setting based on his desired spine muscle activity. Since movements will be slow and not cyclic it is expected the feed-forward control will not result in unstable behavior.

IV. Experiments and Results

The exo-spine, yet without a motor for actuation, was attached to a base to examine the anticipated friction by measuring the forces during up and down movement, and its controllability by verifying the friction's linearity. For testing, and because of the high cable forces, this base was fixed to the forks of a manual forklift, while the cables were attached to a force sensor tight to the forklift base. The exo-spine was bent forward and stretched by lowering and lifting the forks, thereby releasing or pulling the cables. Motions of the exo-spine, obtained using a motion capture system, were recorded simultaneously with the force data.

In order to measure the cable forces at different loads the

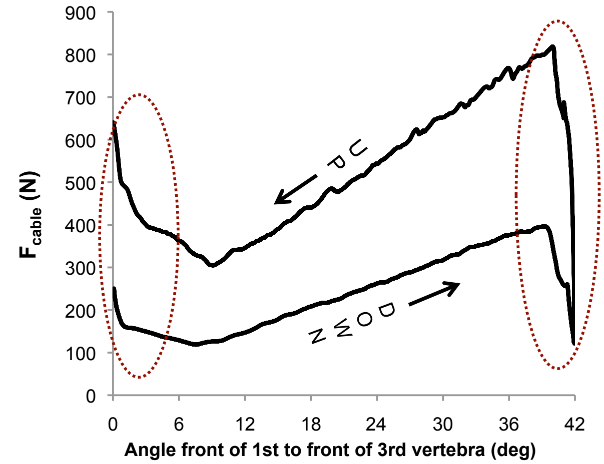


Fig. 5. Graph showing one up-down cycle of the cable forces, F_{cable}, vs. the angle between the 1st and 3rd vertebra and the vertical for a load of 11.2kg. In the left encircled area the neutral position was reached, in the right encircled part the center backup cable stretched.

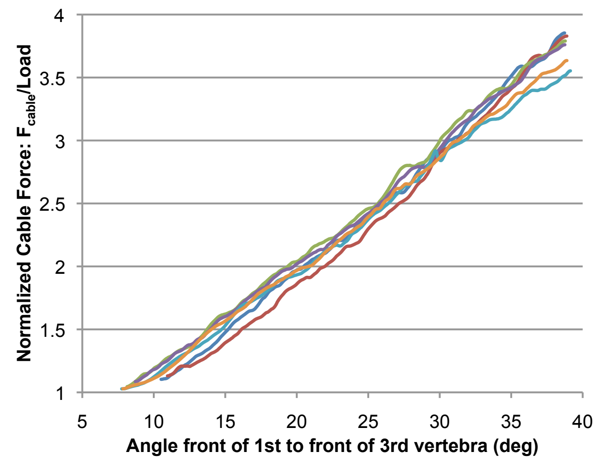


Fig. 6. Normalized cable forces, i.e. F_{cable} divided by the load, during downward bending of the exo-spine for 11.2, 14.3, and 18.0kg loads (twice per value). Important is not so much each individual line as is their closeness, implying linearity of the friction over different loads, as well as their spread, which might hinder effective control.

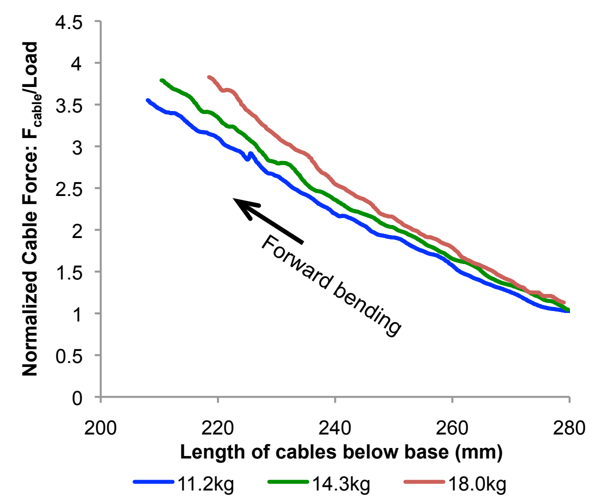


Fig. 7. Length of the cables below the base vs. the normalized cable forces. Increased loads show increased cable lengths due to stretching of the cables.

shoulder blades were locked at zero abduction and arms were attached for hanging weights. Weights were suspended at 37cm from the front of the shoulder blades. A few times, when the exo-spine buckled under loading the load needed to be released for the springs to pull the exo-spine straight again. No other adverse effects from buckling were found.

The ROM were measured to be 64mm abduction of the shoulder blades, 44deg forward bending (as in Fig. 3a), 33deg side bending (Fig. 3b), and 32deg rotation (Fig. 3c).

A. Friction

The exo-spine's friction behavior has been tested to verify the usability of the proposed control method. Fig. 5 shows the cable forces of one up-down cycle for a load of 11kg. At the left encircled part the exo-spine reached the extension limit; the right encircled part occurred when the center safety cable prevented further movement. Comparing the up and down going parts the effect of the friction can be seen.

Applying, e.g., twice as much load results in twice the values for F_{cable}. This can be confirmed from Fig. 6, which shows the normalized cable forces, i.e. F_{cable} divided by the load, during downward bending for 11.2, 14.3, and 18.0kg loads (twice per value) lying close together, indicating that the friction is a linear function of the load. This predictability of the friction is an important prerequisite for feed-forward control. On the other hand, the variability, as seen from the spread of the lines, shows it may not be possible to have accurate control of the assisting force.

In addition, although individual Dyneema fibers can not stretch it was found that, as each cable consists of 12 strands braided together, these strands become squeezed together under tension, which results in a small but significant lengthening of the cable that could distort L_{cable} measurements. This effect can be seen in Fig. 7, where increased loads show increased cable lengths.

V. Discussion

Even though the exo-spine has not yet been combined with HAL there are several indications of how it would perform. Its ROM, for example, can be compared with that of the human spine. Results of various studies are listed in [15] and comparing with these shows that the side bending and rotation ROM are around the average of those reported for the human spine. Although forward bending is about 10deg less, experiments will have to verify the full matching of exoskeleton and wearer before reaching conclusions.

Regarding the shape of the exo-spine, even though it is convex, its sharpest bending point is still at its base. For the human spine this is higher, around the lower thoracic spine. However, most of the exo-spine is not fixed to the wearer's spine and a small gap between the two just above the base can accommodate this difference. Similar effects will be seen for side bending and rotation, in which case differences can be accommodated by a slight S-shape of the exo-spine.

Although the friction is fairly substantial, the device will most not be moving and, owing to the friction, require less motor torque during those times. This also helps to save energy, probably more than offsetting the energy lost to the friction during lifting. As for the likely reduced accuracy of the feed-forward control due to the friction, it is quite likely the wearer will adapt his own back muscle force slightly and even unconsciously to maintain full balance.

From now on the exo-spine will be extended into a simplified full body exoskeleton and worn by several subjects to fully test its performance during lifting. When that is successful a specific control algorithm can be made and tested for SCI patients.

Overall the exo-spine is likely to provide a valuable improvement to the versatility of exoskeleton wearers and consequently to the usability of exoskeletons in general. Its basic structure as well as the usage of fDOF and focus on maximum effectiveness of a minimum amount of actuators could benefit the further design of both exoskeletons and rehabilitation devices such as back supports for SCI patients.

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