MEDARM: a rehabilitation robot with 5DOF at the shoulder complex

Stephen J. Ball, Ian E. Brown and Stephen H. Scott

Abstract— **A key approach for reducing motor impairment and regaining independence after stroke is frequent and repetitive functional training. A number of robotic devices have been developed to assist therapists with the labourious task of providing treatment. Although robotic technology is showing significant potential, its effectiveness for upper limb rehabilitation is limited in part by the inability to make functional reaching movements. A major contributor to this problem is that current robots do not replicate motion of the shoulder girdle despite the fact that the shoulder girdle plays a critical role in stabilizing and orienting the upper limb during activities of daily living. To address this issue, a new adjustable robotic exoskeleton called MEDARM is proposed for motor rehabilitation of the shoulder complex. MEDARM provides independent control of six degrees of freedom (DOF) of the upper limb: two at the sternoclavicular joint, three at the glenohumeral joint and one at the elbow. Its joint axes are optimally arranged to mimic the natural upper-limb workspace while avoiding singular configurations and while maximizing manipulability. This mechanism also permits reduction to planar shoulder/elbow motion in any plane by locking all but the last two joints. Electric motors actuate the joint using a combination of cable and belt transmissions designed to maximize the power-to-weight ratio of the robot while maintaining backdriveability and minimizing inertia. Thus, the robot can provide any level of movement assistance and gravity compensation. This paper describes the proposed technical design for MEDARM.**

Index Terms— **rehabilitation, robot, shoulder girdle, stroke**

I. Introduction

STROKE is a leading cause of disability in Canada, leaving people with motor deficits that limit their ability to perform activities of daily living. Impairments may involve loss of a combination of motor, sensory and/or cognitive functions including weakness, reduced coordination, diminished ability to think, communicate, and make decisions, as well as to feel, see and hear. Currently, more than 300,000 Canadians live with the effects of stroke, and there are 50,000 new occurrences every year. 16,000 Canadians die from stroke every year. Overall, stroke health care costs the Canadian health care system $2.7 billion every year [6]. The costs will only increase as the population continues to age.

It is possible to regain partial or complete motor function through individualized rehabilitation programs. Traditionally, therapists taught compensatory movements such as using redundant joints of the body (i.e. using trunk motion to assist with reaching) or fractionation of movements into several simpler movements [5]. While compensation allows patients to rapidly regain some degree of independence, a strong reliance on compensation promotes learned non-use of the impaired limb which slows or inhibits functional recovery [12]. Current rehabilitation programs tend to focus instead on reducing the degree of permanent disability. Recovery of motor function has recently been linked to motor learning that occurs during repetitive, frequent and intensive movements [16]. This increased sensorimotor activity takes advantage of neural plasticity, which is the ability of adjacent areas of the brain to reorganize and compensate for lost function in other brain regions. It is agreed that exercising and practicing a variety of functional multi-joint movements with the impaired limb is an important part of therapy for stroke patients because it increases their ability to perform activities of daily living [5]. Therefore, typical therapy programs include the use of a variety of techniques such as restraint, gravity compensation, manual guidance, and progressive-resistive exercise.

Although these conventional techniques are effective, a significant drawback is that they require strenuous manual labour and extensive one-on-one attention from therapists. As such, recovery is severely limited by staffing, time, and budget constraints, and it is becoming more difficult to give patients the time and attention that they require for maximum recovery. Even without these limits, therapists can tire or injure easily when manually moving heavy limbs, and with no means to quantitatively record progress, it is a challenge to properly monitor functional ability. Unfortunately, therapists are often forced to resort to shorter, less intense therapy programs that focus on teaching compensatory techniques rather than on recovering motor function [3].

The ever increasing need for motor rehabilitation is straining the capabilities of the health care system. There is no doubt that a more efficient system is required to provide the high quality care that patients need. Robotic systems provide a unique solution, and they offer a number of significant advantages over conventional techniques. Robots can make many precise movements with any level of assistance without getting tired or making mistakes, allowing therapists to focus on treatment planning and progress monitoring. Furthermore, robots can provide a wealth of quantitative measurements for every movement, which could lead to more sensitive functional assessment scores. Other opportunities such as home or group therapy (one therapist with multiple patients) also become feasible using robotic technology.

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There is no question that robots may be ideally suited for rehabilitation and there are a number of projects currently underway that are trying to realize their potential for rehabilitation of the upper limb after stroke [7], [14]. Some existing robots for upper limb rehabilitation and assistance include MIT-MANUS [9], MIME [2], GENTLE/s [11], MULOS [8], T-WREX [15] and ARMin [13] among others.

A significant drawback of current robotic devices is that they cannot match the mobility of the human upper limb. This is particularly true for the shoulder complex because it has a compact arrangement of five major degrees of freedom (DOF): two at the sternoclavicular joint and three at the glenohumeral joint. The glenohumeral joint can be approximated as a ball-and-socket joint and has been replicated in some current devices. However, the shoulder girdle has been neglected, despite its importance in stabilizing and orienting the upper limb. Without direct control at the sternoclavicular joint, it is not possible to prevent compensatory movements, nor is there a means to properly regain strength and coordination of the shoulder girdle. It seems that a next logical step in assessing the usefulness of robotic technology in a rehabilitation setting is to develop a robot that can more closely match the mobility of the human shoulder.

II. Design Objectives

There is no current robotic device that can independently control all five DOF at the shoulder complex. As such, MEDARM (Motorized Exoskelton Device for Advanced Rehabilitation of Motor function) has been designed with shoulder complex control as its fundamental goal. It is intended to be used for assessment in addition to therapy. Figure 1 shows MEDARM for the right upper limb. The design objectives have been described previously [1], but the key goals can be summarized as follows.

- Independent control of five DOF of the shoulder complex (2DOF at the sternoclavicular joint, 3DOF at the glenohumeral joint), and 1DOF at the elbow, with a workspace similar to a typical upper limb workspace.
- Actuators and exoskeleton sized for users from 1.4 m to 2.0 m in height, and weighing up to 115 kg.
- High backdriveability, low mass and inertia to minimize influence on natural motion.
- Allow simplification of the mechanism to 2DOF shoulder/elbow motion in any plane.
- Avoid singular configurations across the workspace.
- Maximize manipulability across the workspace.
- Safe and comfortable for users with motor impairments.
- Quick and easy to set up a patient.

III. MEDARM Specifications

MEDARM is an exoskeleton. An exoskeleton design was chosen because it is the only way to independently control all DOF in the shoulder complex, otherwise the redundancy of the joints would make it impossible to isolate motion of the glenohumeral joint from motion of the shoulder girdle. Another advantage of the exoskeleton design is that grasping a handle is not necessary, leaving the hand available for functional training. The design phase of the MEDARM project is now complete, as shown in Figure 1. The robot is mounted to a support structure, and the user is wheeled into position using a movable chair. There is space for the operator to get beside the exoskeleton during the set-up procedure. MEDARM consists of two main subsystems: the shoulder/elbow mechanism (4DOF to move the upper arm and forearm), and the shoulder girdle mechanism (2DOF to move the glenohumeral joint relative to the torso). The following describes the technical details these subsystems.

Fig. 1. MEDARM system consists of a 6DOF robotic exoskeleton mounted onto a support structure. The motors and electronics are mounted underneath the robot, and external gravity compensation is provided by a motorized vertical cabling system. A movable chair is used to bring the user into alignment with the system.

A. Shoulder/Elbow Mechanism

The shoulder/elbow mechanism (Figure 2a) is a 4DOF mechanism consisting of a 3DOF spherical joint centred at the user's glenohumeral joint and a single rotary joint at the elbow. It is actuated entirely by a cable-drive transmission which is powered by five electric motors located on the base of the system. The overall gear ratio for each of the four joints is 8 so that the robot maintains backdriveability without the need for force/torque sensors. The lightweight mechanism is attached to the lateral side of the user's arm using two adjustable inflatable arm cuffs, which are the only points of physical attachment to the user.

1) Joint Axis Orientation: Glenohumeral motion is achieved using a spherical joint made from three intersecting revolute joint axes. A problem with spherical joints is that it is always possible to reach a singular configuration (where one DOF is lost) by rotating the second joint so that the three axes become coplanar. The order and relative orientation of these three axes was optimized to ensure that the system does not reach singularity within the user's workspace (as specified in [1]), that manipulability is maximized, and that collision with the user or itself does not occur over the entire workspace. The optimization process is described below.

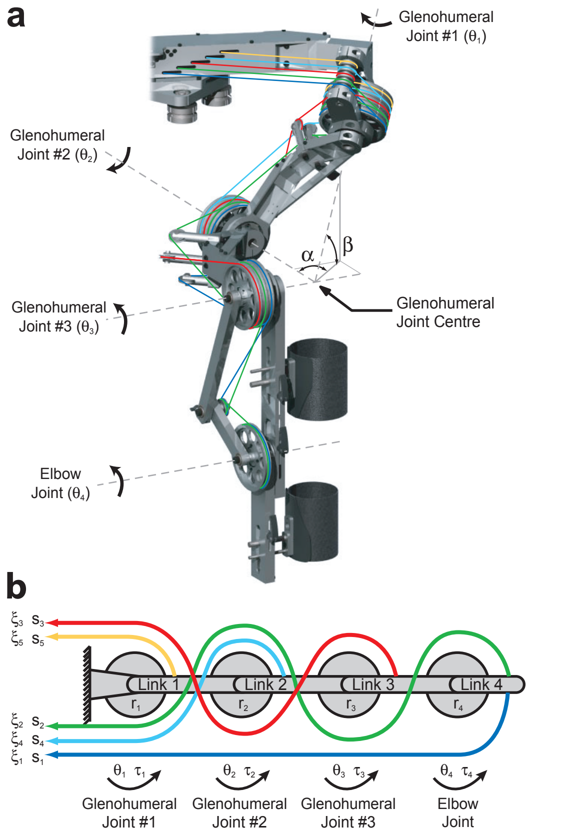

Fig. 2. The shoulder/elbow mechanism is cable-driven. (**a**) A CAD drawing of the mechanism showing the final joint orientations and cable system. (**b**) A simplified planar schematic representation of the optimal cable routing structure. Each of the five cables is denoted by a different colour. Each joint has a separate pulley for each cable that passes by the joint. Symbols s, ξ, r, τ and θ represent cable displacement, cable force, pulley radius, joint torque and joint angle respectively.

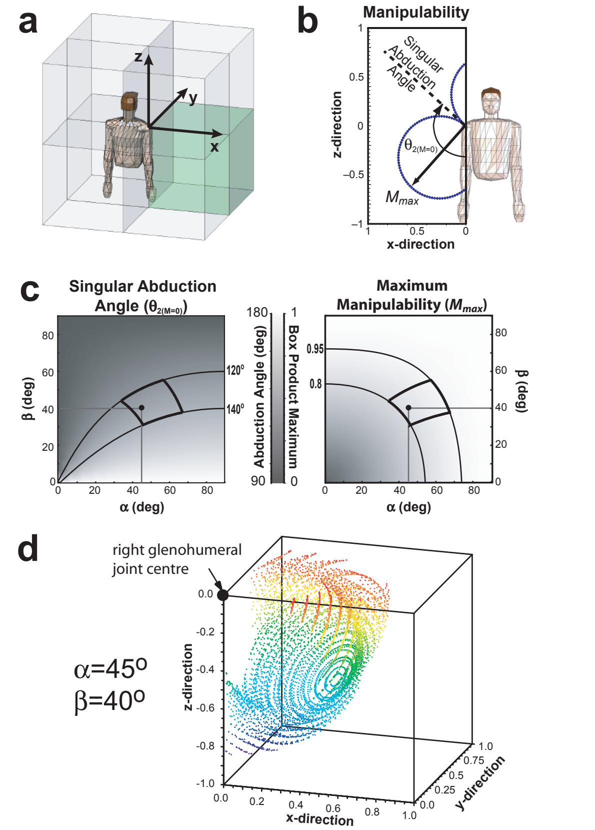

Fig. 3. (**a**) The coordinate frame used for joint orientation optimization calculations. The octant shaded in green approximates the humeral workspace. (**b**) A plot of M for a given combination of α and β as θ_2 is varied to obtain the singular abduction angle ($\theta_{2(M=0)}$) and the maximum manipulability (M_{max}). (**c**) Plots of $\theta_{2(M=0)}$ and M_{max} for all combinations of α and β. The range of α and β combinations that provides a suitable compromise between $\theta_{2(M=0)}$ and M_{max} is shown by contour lines, and the overlap is highlighted. (**d**) M_{max} plotted radially over the workspace. Points closer to the origin are configurations that are closer to singularity.

The first step in choosing the orientations of the axes was to reduce the number of possible configurations by considering the design objectives. In order for the last two joints of the exoskeleton to operate in planar mode, it was necessary to make the last (third) joint axis in the spherical joint parallel to the elbow joint axis (see Figure 2a). With the third joint axis orientation chosen, it was straightforward to determine that the second joint axis should be perpendicular to the third axis (and in the horizontal plane) in order to avoid singularities in the workspace. This configuration also has the added benefit of allowing basic flexion/extension or adduction/abduction motions to be controlled using a single joint axis.

To determine the optimal first joint axis orientation, it was necessary to develop a simple iterative procedure to calculate the box product, M, at each configuration in the workspace. The box product is defined as:

$$M = z_1 \times (z_2 \times z_3) \tag{1}$$

where z_i are the unit vectors corresponding to the joint axes. When $M = 1$, the joint axes are orthogonal and manipulability is maximized. When $M = 0$, the joint axis are coplanar and a degree of freedom is lost (i.e. singular configuration). The procedure is summarized as follows, and step-by-step results are shown in Figure 3:

1) The orientation of the first joint axis was defined relative to the second joint axis in terms of two variables (α and β, as shown in Figure 2a).
2) With θ_1 and θ_3 fixed, manipulability (M) was calculated for a combination of α and β as θ_2 was varied (corresponds to abduction, as shown in Figure 3b).
3) The singular abduction angle ($\theta_{2(M=0)}$) and the maximum manipulability (M_{max}) were calculated and plotted for all combinations of α and β (Figure 3c).
4) A range of α and β combinations that reached a compromise between high M_{max} and large $\theta_{2(M=0)}$ was revealed (i.e. $M_{max} > 0.8$ and $\theta_{2(M=0)} > 120°$). The following iterative procedure was then used to select a combination of α and β within this range.
5) M was calculated for the spherical joint workspace for a given α and β (all three joints varied across their ranges of motion, as shown in Figure 3d).
6) If any points were within 15° of singularity ($M < 0.3$) or if the exoskeleton could collide with the user,

the process was repeated from step 5 using a different combination of α and β.

7) The process was repeated until there were no singularities in the workspace, and the manipulability was as high as possible over the workspace.

The angles that provided the optimal spherical joint axes arrangement are $\alpha = 45°$ and $\beta = 40°$. The maximum manipulability is 0.85 and averages 0.75 across the workspace.

2) Cable-Drive System: The joints of the shoulder/elbow mechanism are actuated by electric motors with an open-ended cable drive transmission. Cable-driven mechanisms have a high power-to-weight ratio because all the motors can be placed on the fixed base of the system. This significantly reduces the size, mass and inertial properties of the robot, and helps to reduce the motor torque output requirements. Open-ended cable systems distribute loads across several cables, which also reduces the the actuator requirements. Overall the mechanism becomes more transparent to the user.

Additional transformations are required to control open-ended cable-drive systems [17]. This is because the number of joints needing control (n) is less than the number of actuators (m). Cable systems can apply force through tension only, so it is necessary to have an antagonistic cable routing scheme for motion capability in both rotational directions at each joint. As such, a minimum of $n + 1$ cables are necessary for complete control of n joints. It is necessary to have a positive tension in all cables at all times to prevent the cables from becoming slack. Furthermore, since the cables are routed along the entire length of the mechanism through a series of pulleys, their motion affects multiple joints, allowing loads at the joints to be distributed among the actuators. Ultimately, joint angles and torques are related to the length displacement and the forces in the cables.

The choice of cable routing scheme has a significant effect on the performance of the device. In fact, for this 4DOF system ($n = 4$) actuated by five motors ($m = 5$), there are 11 possible unique cable routing structures, all of which can be described in matrix form. This structure matrix can be analyzed to obtain quantitative measures related to efficiency and actuator torque requirements [10]. Figure 2b illustrates the optimal routing scheme which has minimized antagonism between cables and hence the most even distribution of forces across the cables, and also has the lowest peak forces.

3) Joint Design: Each joint of the exoskeleton requires a low friction bearing system that provides rigidity against all forces and non-axial moments. In addition to withstanding non-axial gravitational and inertial moments during motion, the joints must withstand substantial non-axial moments resulting from forces applied by the cables and pulleys. Four-point contact bearings are highly resistant to these moments and therefore need not be used in pairs. Use of four-point contact bearings in MEDARM has resulted in a thin and lightweight exoskeleton.

B. Shoulder Girdle Mechanism

The shoulder girdle mechanism (Figure 4) provides 2DOF about the sternoclavicular joint centre: elevation/depression and protraction/retraction. The entire mechanism is located behind the user, and there is an adjustment to account for users of different size. The mechanism supports the complete shoulder/elbow system including the user's arm, and as a result must be structurally strong.

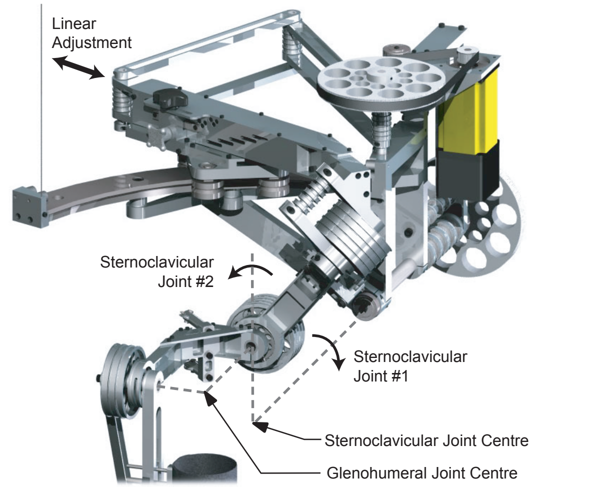

Fig. 4. A CAD drawing of the shoulder girdle mechanism. The two joint axes intersect at the user's sternoclavicular joint. The second joint is a translation along a curved track, producing a rotation about the vertical axis. The joint is driven by a hinged linkage system. A single linear adjustment shifts the entire shoulder/elbow mechanism to align the spherical joint with the user's glenohumeral joint centre.

The first joint axis is fixed to the base structure behind the user, with its axis pointing forward in the horizontal plane. It is a conventional rotary joint that provides elevation/depression motion. The second joint axis is vertically aligned, and intersects the first joint axis through the user's sternoclavicular joint centre, allowing protraction/retraction motion. It is not a typical rotary joint; it is a curved track system on which a carriage travels. The low-friction carriage supports the entire cable drive system and is driven by a hinged linkage system. The resulting mechanism operates like a 4-bar linkage without requiring any structural elements near the user's sternoclavicular joint (see Figure 4). Both joints are driven by electric motors with timing belts, and operate with gear ratios of 5 and 6.25 respectively.

The benefits of this track system are significant. First, it facilitates placing equipment behind the user rather than above their head, which is safer and more comfortable for the user, and also easier for the operator to set up. Second, the hinged driving linkage doubles as a routing system for the cables from the shoulder/elbow mechanism by guiding them through to the base of the robot without any non-linear changes in cable length. Any change in cable length as a result of shoulder girdle motion is easily accounted for in the cable length transformations.

The weight of this mechanism is substantial, and puts high static torque requirements on the first shoulder girdle joint. To assist the motor at this joint, an external gravity compensation system is employed. A vertical motorized cable is mounted directly above the end of the curved track (see Figure 1), and applies a vertical force on the track to offset the gravitational forces on the MEDARM.

C. User Attachment and Alignment

To function correctly, the exoskeleton must be aligned with the sternoclavicular and glenohumeral joints of the user, and adjusted to fit arms of different lengths. A harness attaches the user's torso to a moveable chair which provides the three translational adjustments necessary to align the user's sternoclavicular joint with the mechanism's fixed shoulder girdle joint centre. Once aligned, the chair is locked to the main structure.

The mechanism must next be aligned with the user's glenohumeral joint centre. As before, three spatial adjustments are required. However, this can be achieved by a single manual linear adjustment because the redundancy of the exoskeleton structure can be used to make the remaining alignments. This linear adjustment shifts the cable-drive system relative to the carriage in the direction approximately aligned with the horizontal projection of the clavicle (see Figure 4) and is then clamped to the carriage. Thus modifying the position of the mechanism's glenohumeral joint centre is achieved through the linear adjustment and the 2DOF provided by the shoulder girdle. The 3DOF spherical joint of the shoulder/elbow mechanism automatically compensates by rotating until the mechanism is properly aligned with the user's limb.

This adjustment scheme has the benefit of simplifying the structure of the shoulder girdle joint, and also the set-up procedure. Otherwise, three consecutive translational adjustments would be required, making the system significantly larger, heavier and more complicated. Relying on the shoulder/elbow mechanism to compensate for the adjustment tends to push the shoulder joint away from its optimal configuration, decreasing the range of motion of the mechanism in some directions. However, the adjustment range is typically small (2° or 3° at most), so the singularities and manipulability of the mechanism will not be significantly altered. Another issue that arises when adjusting a cable-drive system is that it is necessary to maintain tension in the cables at all times. Adjusting the link length must not change the cable length, otherwise tension would be lost. Routing the cables along the hinged driving linkage ensures that the cable length does not change and that tension is maintained.

The exoskeleton system attaches to the user in two places: the upper arm and the forearm (Figure 5a). These attachments are to keep the exoskeleton aligned with the limb at all times. The proposed design is to strap the limb into a rigid half-cylindrical trough using an inflatable Velcro strap similar to a blood pressure cuff. Once strapped in, the cuff is inflated to provide a secure fit that is customized to the user. On the lateral side of the cuffs is a single rigid connection to the exoskeleton structure. The cuff will be attached to the subject before connecting to the exoskeleton, which is easier for the operator, and more comfortable for the user. An important difference from many previous arm cuff designs is that the arm cuff does not have a fixed size cuff through which the user must put their arm. This allows simpler set up, and also is compatible with a larger variety of arm sizes.

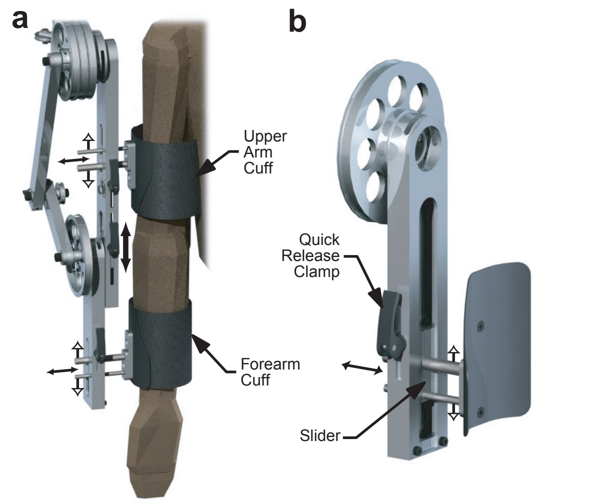

Fig. 5. (**a**) A CAD drawing illustrating the arm cuff attachments and adjustments. Each cuff has two translational adjustments to correctly align the limb segments relative to the mechanism structure: perpendicular to the link (small arrows) and parallel to the link (hollow arrows). A fifth adjustment (large arrow) moves the location of the elbow joint to change the length of the upper limb link. (**b**) A close-up of the cuff attachment, showing the quick-release clamp.

A total of five adjustments are required to ensure that the user's arm is properly aligned with the exoskeleton (see Figure 5a). Each cuff is adjustable along the length of the exoskeleton (for limbs of different length) and perpendicular to the exoskeleton (for limbs of different width). The cuff is attached by inserting it into a slider which can move freely along the exoskeleton. A single quick-release clamp (similar to those used to clamp bicycle components) simultaneously clamps the cuff to the slider and the slider to the exoskeleton (see Figure 5b). To accommodate users with different arm lengths, a similar slider and clamp is used to locate the elbow joint along the upper arm link. A passive hinged guide was added to the upper arm link of the robot to ensure that tension is not lost when adjusting the arm length.

Exoskeleton type devices always require more set up time than their end-effector type counterparts. However, given its mobility and adjustability, MEDARM has a relatively simple set up procedure. In fact, once the chair is locked in place, only four clamps are required to secure all eight adjustments. This will keep set up time to a minimum, allowing the user to receive a longer therapy session.

IV. Dynamic Model and Simulation

To make appropriate choices for the eight electric motors required to actuate MEDARM, a dynamic model of the exoskeleton and the human limb has been created in MATLAB based on the robot toolbox [4]. The model was also used to specify a number of other design parameters including bearing strength, joint gear ratios, and cable load capacity.

The model takes the form of a standard rigid-body manipulator, and assumes that the cable dynamics are not

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TABLE I

MAXIMUM TORQUE OUTPUT FOR EACH JOINT.

Motor	Static Torque (Nm)	Peak Torque (Nm)
Shoulder Girdle #1	±24	±73
Shoulder Girdle #2	±30	±91
Glenohumeral #1	+39, −26	±60
Glenohumeral #2	+39, −26	±60
Glenohumeral #3	+39, −26	±60
Elbow	±13	+40, −30

significant. Dynamic parameters of the exoskeleton including lengths, masses and inertial properties are estimates from CAD drawings. The same properties of the human upper limb were calculated from anthropometric data tables based on user height and weight [18] and are fully integrated into the model. The model was adapted to account for the external gravity compensation system, and includes estimates of viscous and static friction. Given a trajectory for each joint, the model calculates the joint torques required to achieve that motion. The cable forces required to generate these joint torques are then calculated using the torque resolver technique [10]. The final output is the torque outputs for all eight motors, the force in each cable, and all forces and non-axial moments at each joint.

To get an estimate of the peak dynamic motor torques for non-contact applications, the model was used to simulate various reaching movements with a peak end-point velocity of 1.0 m/s. Anthropometric limb measurements were chosen to meet the maximum design requirements. Movements included single joint movements through the full range of motion of each joint, and a range of typical multi-joint reaching movements such as reaching towards the face or chest from a relaxed position. The most demanding positions for the exoskeleton system in terms of static torque requirements are those in which the arm is raised to the horizontal plane with the elbow fully extended. The gravitational component of the joint torques is the most significant contribution, and produces the largest stresses on the motors in static situations, so each position was held for one second to facilitate measurements of peak static torque.

Motors and gear ratios were selected based on the results of these simulations. The motors have built-in high resolution encoders capable of measuring joint angle in increments of 0.003°. Each motor also incorporates an electric brake to guarantee that the mechanism will not collapse during a power failure. The brakes also ensure that the cables remain in tension when the power is turned off. The simulations also enabled selection of a braided stainless steel cable of appropriate size, and also joint bearings with sufficient load capabilities. The overall torque capabilities of each joint of the exoskeleton are shown in Table I, and are a result of the limits of both the motors and the cable strength.

V. CONCLUSIONS AND FUTURE WORK

This paper describes the design of a new robot for assessment and rehabilitation of upper-limb motor function after stroke. MEDARM is designed to be a versatile machine with the potential to improve motor function of the proximal upper limb. It is hoped that MEDARM will assist in prevention of compensatory movements, while encouraging more natural coordination for stroke patients of any degree of motor impairment. The robot provides a large range of adjustments, so it can easily accommodate users of varying shape and size. MEDARM's main advantage, however, is that it can independently monitor and control all five of the main DOF of the shoulder complex.

Efforts are currently focused on construction and testing of a simpler 3DOF planar version of MEDARM to test out the main principles of operation of the MEDARM design including the curved track and cable-drive transmission.

REFERENCES

[1] S. J. Ball, I. E. Brown, and S. H. Scott, "Designing a robotic exoskeleton for shoulder complex rehabilitation," in *30th Canadian Medical and Biological Engineering Conference (CMBEC30)*, Submitted for Publication, 2007.

[2] C. G. Burgar, P. S. Lum, P. C. Shor, and H. F. Machiel Van der Loos, "Development of robots for rehabilitation therapy: the Palo Alto VA/Stanford experience," *J. Rehabil. Res. Dev.*, vol. 37, no. 6, pp. 663–673, 2000.

[3] M. C. Cirstea and M. F. Levin, "Compensatory strategies for reaching in stroke," *Brain*, vol. 123, no. 5, pp. 940–953, 2000.

[4] P. I. Corke, "A robotics toolbox for MATLAB," *IEEE Robot. Automat. Mag.*, vol. 3, no. 1, pp. 24–32, 1996.

[5] B. H. Dobkin, "Strategies for stroke rehabilitation," *Lancet Neurol.*, vol. 3, no. 9, pp. 528–536, 2004.

[6] Heart and Stroke Foundation of Canada. (2007) Stroke statistics. [Online]. Available: http://www.heartandstroke.ca

[7] S. Hesse, H. Schmidt, C. Werner, and A. Bardeleben, "Upper and lower extremity robotic devices for rehabilitation and for studying motor control," *Curr. Opin. Neurol.*, vol. 16, no. 6, pp. 705–710, 2003.

[8] G. R. Johnson, D. A. Carus, G. Parrini, M. S. Scattareggia, and R. Valeggi, "The design of a five-degree-of-freedom powered orthosis for the upper limb," *Proc. Inst. Mech. Eng [H.]*, vol. 215, no. 3, pp. 275–284, 2001.

[9] H. I. Krebs, N. Hogan, M. L. Aisen, and B. T. Volpe, "Robot-aided neurorehabilitation," *IEEE Trans. Rehabil. Eng*, vol. 6, no. 1, pp. 75–87, 1998.

[10] J. J. Lee, "Tendon-driven manipulators: Analysis, synthesis, and control," Ph.D. Thesis, University of Maryland, 1991.

[11] R. Loureiro, F. Amirabdollahian, M. Topping, B. Driessen, and W. Harwin, "Upper limb robot mediated stroke therapy - GENTLE/s approach," *Autonomous Robots*, vol. 15, no. 1, pp. 35–51, 2003.

[12] S. M. Michaelsen, R. Dannenbaum, and M. F. Levin, "Task-specific training with trunk restraint on arm recovery in stroke: randomized control trial," *Stroke*, vol. 37, no. 1, pp. 186–192, 2006.

[13] M. Mihelj, T. Nef, and R. Riener, "ARMin - toward a six DoF upper limb rehabilitation robot," in *Proc. IEEE/RAS-EMBS International Conference on Biomedical Robots and Biomechatronics (BIOROB'06)*, 2006, pp. 1154–1159.

[14] D. J. Reinkensmeyer, J. L. Emken, and S. C. Cramer, "Robotics, motor learning, and neurologic recovery," *Annu. Rev. Biomed. Eng*, vol. 6, pp. 497–525, 2004.

[15] R. Sanchez, D. Reinkensmeyer, P. Shah, J. Liu, S. Rao, R. Smith, S. Cramer, T. Rahman, and J. Bobrow, "Monitoring functional arm movement for home-based therapy after stroke," in *Proc. IEEE Annual Conference of Engineering in Medicine and Biology Society (EMBC'04)*, vol. 2, 2004, pp. 4787–4790.

[16] R. Teasell, N. Foley, J. Jutai, T. Doherty, J. Bitensky, S. Bhogal, and M. Speechley, "Stroke rehabiliation evidence-based review: Upper extremity interventions," Monograph on the Internet, 2003.

[17] L.-W. Tsai, *Robot Analysis: The Mechanics of Serial and Parallel Manipulators*. New York, U.S.A.: John Wiley & Sons, Inc., 1999.

[18] D. A. Winter, *Biomechanics and Motor Control of Human Movement*, 2nd ed. New York, U.S.A: John Wiley & Sons, Inc., 1990.