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Quantifying Anti-Gravity Torques in the Design of a Powered Exoskeleton

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***Abstract*— Designing an upper extremity exoskeleton for people with arm weakness requires knowledge of the passive and active residual force capabilities of users. This paper experimentally measures the passive gravitational torques of 3 groups of subjects: able-bodied adults, able bodied children, and children with neurological disabilities. The experiment involves moving the arm to various positions in the sagittal plane and measuring the gravitational force at the wrist. This force is then converted to static gravitational torques at the elbow and shoulder. Data are compared between look-up table data based on anthropometry and empirical data. Results show that the look-up torques deviate from experimentally measured torques as the arm reaches up and down. This experiment informs designers of Upper Limb orthoses on the contribution of passive human joint torques.**

I. INTRODUCTION

The larger project involves development of a powered upper extremity orthosis that assists people with muscular weakness to perform activities of daily living. A passive arm exoskeleton has been developed and commercialized by this group [3]. The device is called the Wilmington Robotic Exoskeleton (WREX), which is a gravity balanced upper limb orthosis for children with muscular weakness such as musclular dystrophy and spinal musclular atrophy. The WREX has four degrees of freedom to allow full range of motion, and is assisted by gravity balancing elastic bands [3]. Typically, the WREX is attached to a wheelchair or to a body jacket.

An external power source has been added to the WREX to overcome two current problems [2]:

1) A child with muscular weakness often has difficulty in raising his arm above his head, even with the WREX.

2) The child cannot lift a substantial weight, because the device only balances the mass of the child's arm.

A model of the human arm is required in order to control the device. Of particular importance is the proper characterization of the passive joint torques at the human elbow and shoulder. These data are not available for people with neuromuscular disease. This paper summarizes a series of experiments to measure the passive static joint torques through the range of vertical motion.

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Some previous work has been done to characterize strength of children as well as the specific diseases included in this study. Sunnegrdh investigated the strenght of normal children ages 8 and 13 [4]. Mathur et al. studied the time dependent linear decrease in muscular strength of subjects with Duchenne Muscular Dystrophy [1], however this is not sufficient in modeling a subject for control of an upper limb exoskeleton. Therefore, this study was conducted to obtain a preliminary understanding of differences between upper limb properties between adults, healthy children, and children with specific disabilties, namely spinal muscular atrophy, arthrogryposis, and muscular dystrophy. The end goal is to obtain a robust model that can be used to control a powered assistive device.

The controller proposed for our powered orthosis uses residual force input from the user as a measure of their intention. The force sensor picks up inertial, Coriolis, viscous, elastic, and gravitational forces as well as the voluntary force of the user. In slow movements, typical of disabled children, the passive terms - gravitational and elastic - are much larger than the other force components. However, the ratio of voluntary to gravitation force is very small for weak individuals therefore it becomes important to accurately characterize the passive forces (gravitational and passive joint resistances) in order to better measure the voluntary component. Once a general pattern of passive joint torques for people with neuromuscular disabilities is determined it can be used to modify the torques derived from the look up table. The look up table is based on a person's height and weight and only applies to people without disabilities.

II. METHOD

A. Human Model

A model of the user's arm is needed. The initial model for the human arm was a two link lumped mass model with pin joints shown in Fig. 1. The model was limited in the vertical plane - the primary direction of assistance from the WREX. Values for segment mass and center of mass were obtained from anthropomorphic tables based on the subjects height, weight, and limb segment lengths [5]. However, initial testing showed that this model was not accurate compared to experimental values. Therefore the shoulder and elbow joint torques in the vertical plane were measured to quantify that difference.

B. Experimental Protocol

For each subject several body measurements were taken, including height, weight, upper arm length, and lower arm

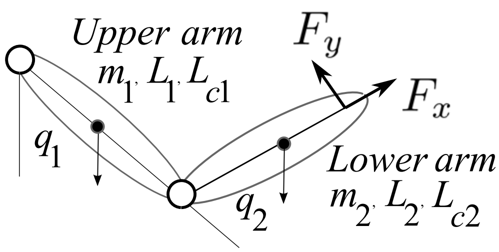

Fig. 1. Model of upper limb as two rigid links with given variables

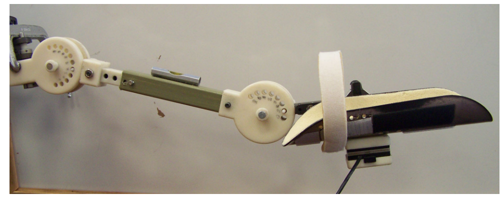

Fig. 2. Measuring device with lockable joints at the shoulder and elbow with a force sensor attached to the arm trough

length. The measuring device was adjusted to fit the subject, who sat in a chair or in a powered-wheelchair. The shoulder joint of the device was aligned with the shoulder joint of the subject in a horizontal position. The human shoulder joint has a vertical displacement throughout its range of motion, which causes misalignment between the device and the subject. The effects of this misalignment would be recorded in the force sensor. The subject's forearm was attached to the trough using a wrist splint with Velcro straps. The subject's dominant arm was measured, since the WREX is typically used on this side. The device was locked at a maximum of 9 shoulder joint positions from 10 to 150 degrees from the vertical in 20 degree increments and 8 elbow joint positions from 0 to 140 degrees relative to the upper arm in 20 degree increments. For each locked position, the subject was instructed to relax his/her arm. The device would fully support the passive moments of the arm. This would ensure that only passive human joint torques would be recorded. The number of positions was reduced to remain within the comfortable range of the subject. All of the subjects with disabilities had a significantly reduced range of motion. At each position, a reading from the force sensor was taken. This reading included two forces along the x and y axis and a moment about the z axis. The subject was instructed to relax to prevent misreadings from voluntary muscle activation and were reminded to relax periodically throughout the experiment. The procedure was granted approval by the Internal Review Board (IRB). Consent forms were obtained from all subjects over the age of 18 yrs. Assent forms were obtained from subjects less than 18 yrs along with parental permission.

The measuring device is shown in Fig. 2. It has two adjustable links with lockable shoulder and elbow joints in 20 degree increments. An arm trough is connected to a 6 axis force/torque sensor (ATI, Apex, NC). Only the 2 force directions and 1 moment direction that act in the vertical plane were recorded. A wrist splint attaches the subjects arm to the device. The force readings are transformed into joint torques at the shoulder and elbow using the Jacobian. Three groups were included in this study: (1) 5 Adults, age 19-50 with no upper limb disability (2) 4 Children, age 13-18 with no upper limb disability (3) 4 Children, age 13-18 with either arthrogryposis, muscular dystrophy, or spinal muscular atrophy.

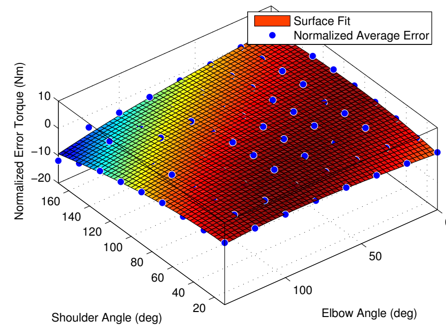

Fig. 3. Average Adult error torque at the shoulder normalized to average adult mass with third degree polynomial fit

III. RESULTS

The expected forces from the anthropomorphic table and measured forces for each subject were transformed into joint torques at the shoulder and elbow joint using the Jacobian. For the adults and healthy children, the differences between the table values and measured values were calculated. The differences were normalized to the average weight of the group and plotted. It was found that a third degree polynomial adequately fit the subsequent averaged, normalized data shown in Figs. 3 and 4 for the adults and Figs. 5 and 6 for healthy children.

The third degree polynomial in 1 was fit to the normalized

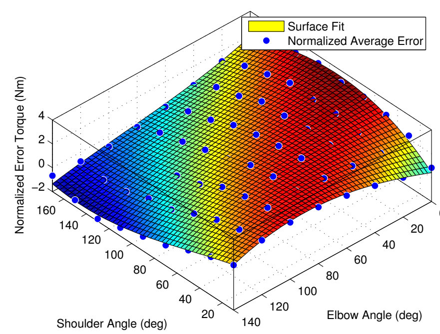

Fig. 4. Average Adult error torque at the elbow normalized to average adult mass with third degree polynomial fit

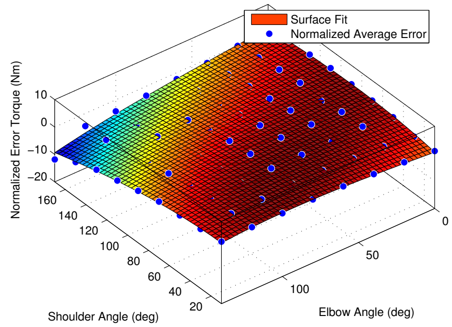

Fig. 5. Average healthy child error torque at the shoulder normalized to average mass of the group with third degree polynomial fit

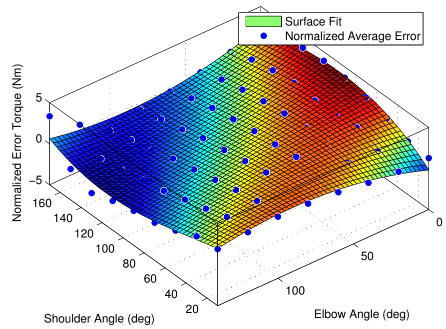

Fig. 6. Average healthy childe error torque at the elbow normalized to average mass of the group with third degree polynomial fit

data for both the adults and healthy children:

$$\begin{aligned} f(x,y) &= p_{00}+p_{10}x+p_{01}y+p_{20}x^2 \\ &\quad +p_{11}xy+p_{02}y^2+p_{30}x^3 \\ &\quad +p_{21}x^2y+p_{12}xy^2+p_{03}y^3 \end{aligned} \tag{1}$$

where x is the shoulder angle and y is the elbow angle in degrees.

The coefficients in 1 for each group and joint are:

Adult Shoulder:

$p_{00}:-1.8, p_{10}:0.067, p_{01}:0.089, p_{20}:-0.00025,$
$p_{11}:-0.00039, p_{02}:-0.00031, p_{30}:-4.8\times10^{-7},$
$p_{21}:-3.0\times10^{-8}, p_{12}:-8.2\times10^{-7}, p_{03}:-2.1\times10^{-7}$

Adult Elbow:

$p_{00}:-0.46, p_{10}:0.094, p_{01}:0.058, p_{20}:-0.00066,$
$p_{11}:-0.0012, p_{02}:-0.00014, p_{30}:1.1\times10^{-6},$
$p_{21}:3.6\times10^{-6}, p_{12}:2.4\times10^{-6}, p_{03}:-1.0\times10^{-6}$

Healthy Children Shoulder:

$p_{00}:-0.24, p_{10}:0.086, p_{01}:0.049, p_{20}:-0.00062,$
$p_{11}:-0.00065, p_{02}:-1.0\times10^{-5}, p_{30}:1.0e{-6},$
$p_{21}:-4.2\times10^{-8}, p_{12}:7.7e{-7}, p_{03}:-1.5\times10^{-6}$

Healthy Children Elbow:

$p_{00}:-1.3, p_{10}:0.12, p_{01}:0.082, p_{20}:-0.00093,$
$p_{11}:-0.0017, p_{02}:-0.00040, p_{30}:1.8\times10^{-6},$
$p_{21}:5.3\times10^{-6}, p_{12}:4.5\times10^{-6}, p_{03}:2.9\times10^{-8}$

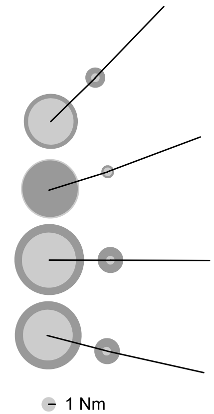

Fig. 7. Passive shoulder and elbow torque in various positions from the model and experimental data for a male, 16 year old with Spinal Muscular Atrophy

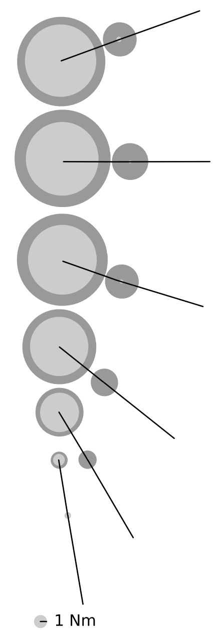

Fig. 8. Passive shoulder and elbow torque in various positions from the model and experimental data for a female, 13 year old with Arthrogryposis

The data of the disabled children was not averaged together, because of the limited number of subjects with similar disabilities. Instead, the joint torques are represented graphically for several positions throughout each subjects comfortable range of motion shown in Figs. 7 to 10. The size of the circle is scaled to the joint torque. The dark gray represents the table values. The light gray represents the measured values.

IV. DISCUSSION

The differences between the anthropomorphic values and the measured torques come from joint misalignment, joint stiffness, and differences between table values and true values. The goal of this project was to lump these errors together in order to improve the human model at the joint level. The

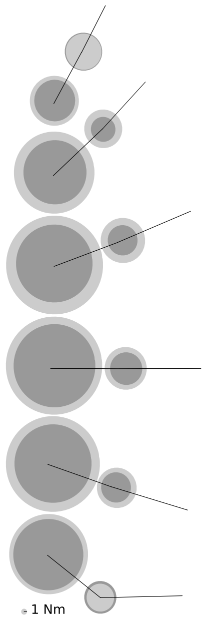

Fig. 9. Passive shoulder and elbow torque in various positions from the model and experimental data for a male, 14 year old with Muscular Dystrophy

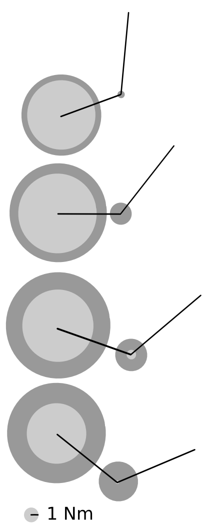

Fig. 10. Passive shoulder and elbow torque in various positions from the model and experimental data for a male, 18 year old with Muscular Dystrophy

fitted curves in Figs. 7 to 10 followed a similar shape between healthy children and adult groups. This indicates that the sources of errors is relatively consistent across age groups. There are two things to note in comparing the values of the coefficients of the surface fit polynomials between the adults and healthy children. First, most of the corresponding coefficients are within the same order of magnitude. This is expected because the size and weight of the healthy children were within the same order of magnitude of the adults. Second, for both groups, there is a general descending magnitude in the coefficients corresponding to higher order terms. This is also expected since the square and cube terms of the angle becomes large at the extremes. However, it should be noted the the coefficients have no direct correlation to physical properties of the subject. The given third degree polynomial with coefficients can be scaled to a specific subjects weight and used to augment a two link model from anthropomorphic table data.

The pattern of the measured joint torques of children with disabilities was similar to healthy children, indicating that gravity is also the dominant component of passive torque in the disabled children as well. One major difference was that all of the disabled children had a limited range of motion when compared to normal children. In the children with disabilities group, the errors between anthropomorphic table and actual data varies depending on disability. In three of the subjects, the measured torque values were significantly less than expected especially at the elbow joint. This was in part due to disabilities that significantly decrease arm mass, such as spinal muscular atrophy. However, one subject had slightly higher measured values shown in Fig. 9. Therefore, subject specific models need to be created for each subject depending on each medical condition.

V. CONCLUSION

In forming a model of a human arm, measurements of the passive joint torques in the vertical plane showed that a two link model is inadequate to describe the human arm. It was found that normal adults and children have a similar shape in torque differences that can be represented by a third degree polynomial. The children with disabilities in this study did not have similar curves and could not be averaged across disabilities. A subject specific model is needed. The next step will be to use this model with an attached force sensor to a powered WREX to measure and amplify the intention of the user.

VI. ACKNOWLEDGMENTS

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