Modeling and Simulating Lifting Task of Below-Knee Amputees

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Abstract. Lifting is a common activity to below-knee amputees (BKA) in occupational and living occasions. Appropriate lifting posture is crucial to physical safety and health to those BKAs. Often healthy parts of BKAs might be hurt due to extra and asymmetric force exertion compensating for deficiency of disabled body parts. To prevent further hurt, a validated biomechanical model describing lifting is essential to analyze lifting behavior of those handicapped. In this study, twelve BKAs were recruited to lift 45 N weights from the floor. Subjects are asked to lift three levels of weights (0 N, 30 N, 60 N) by two postures: squat lifting and stoop lifting. Twelve non-BKAs were recruited as comparison group to study the variance caused by disability. Calculated forces based on Anybody were compared with EMG signals of body parts on spine and thigh. A framework of three-level constraints models were applied to adjust the difference between calculated forces and EMGs and the results validate the model.

Keywords: Below-knee · Amputees · Lifting · Modeling

1 Introduction

Among various work-related activities, lifting, awkward posture, and heavy physical work have been indicated to have strong relationship with lumbar musculoskeletal disorders (MSDs) [1]. Combination of lifting with lateral bending or twisting has been identified as a frequent cause of back injury in the workplace [2, 3]. Search for an appropriate lifting technique has thus attracted considerate attention due to high risk of injury. Compression force limits have been recommended [4] for safer material handling maneuvers based on the premise that excessive compression loads could cause injury.

Role of lifting in low-back injuries is well recognized in the literature as described before. Despite researches in low-back injuries, literatures on safer lifting methods are still controversial. Among all methods lifting low-lying objects, squat lifting (i.e., knee bent and back straight) and stoop lifting (i.e., knee straight and back bent) are two main methods, and the former is considered safer than the latter, for squat lifting brings the load closer to the body center compared to stoop lifting and reduce the extra demand on back muscles to counterbalance additional moments. Few further researches comparing two methods from a biomechanical view and the importance of selection between two

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postures are downplayed [5]. And many workers, without any occupational safety guidelines, prefer the stoop lift over squat lift. There are other conclusions that there is an increased physiological cost and more muscle fatigue are developed when squat lifting [6]. The inability to accurately determine the loads on trunk active and passive components as well as the system stability margin appears as a critical hindrance towards the development of ergonomics guidelines for the design of safer lifting tasks. Evidently, an improved assessment of risk of injury depends on a more accurate estimation of the load partitioning in human trunk.

As a special population, it is of interest to determine if it is feasible to apply a discount factor in terms of the differences in the mass selected by disabled workers in comparison to non-disabled workers for the varying task. There have been many studies conducted involving manual materials handling (specifically lifting). However, not many tools and researches involve a separation between the ordinary and special populations. Wright and Mital have done two studies involving lifting and carrying [7]. These studies were done in order to recreate the situations presented by Snook and Ciriello [8] to investigate the muscle strengths used by an older population when performing routine activities in industrial and home environments. Chen observed older (50 years and older) and younger (20–30 years old) manual materials handlers and made comparative analysis on the maximum acceptable lifting masses differed between young and older female workers, potential age-related differences in kinematic lifting strategies, grip strength, ratings of perceived exertion [9].

The goal of ergonomics is to design the job to fit the individual performing the job. This includes the worker's mental and physical capabilities, limitations and tolerances [10]. There are many different ways in order to measure the various tool and safety thresholds for workers. Some examples of tools used to evaluate occupations are rapid upper limb assessments (RULA), National Institute for Occupational Safety and Health (NIOSH), Snook and Ciriello tables, and 3D Static Strength Prediction Program (3DSSPP). It was also of interest to determine if the biomechanical model induced from observations on non-disabled people still work for handicapped. Data collected in this study are expected to increase biomechanical data activating biomechanical lifting model assisting the development of evidence-based disability-specific guidelines for safely designing manual materials handling tasks.

There are many approaches to study the biomechanical model of the spine and the force-exertion muscles. Kinematic-based approach is quite useful to compute muscle forces and spinal loads at different spinal levels in static lifting activities involving flexion of trunk and lower limbs. The study is to validate a novel kinematic-based approach applied in El Rich's research in the case that the BKAs maneuver lifting tasks with two methods: squats lifting and stoop lifting [11]. The kinematic-approach was conducted within a framework of three-level model, which integrate the two kinds of constraints: task constraints and handicapped function constraints, and thus model the specific behavior of the physical handicapped in the virtual environment. Based on 3 levels of constraints, the model predicts the optimization of strength and torque under physical and dynamic constraints of physical disability. The simulated results were compared with experiment results and validity of three-level model was validated.

2 Method

Twelve male BKAs were recruited to participate in the experiment with another group of non-BKAs with no recent back complications volunteered for the study after signing an informed consent form. The mean (± S.D.) age, body weight and mass of the BKA group were 36 ± 4 years, 169 ± 5 cm and 65 ± 9 kg, while the mean of non-BKAs were 34 ± 3 years, 172 ± 6 cm and 68 ± 5 kg. The experiment was set as Fig. 1, where the kinetic and muscle EMG signals were collected for the biomechanical analysis of two different lifting methods. A six-camera VICON system (VICON, UK) was employed to collect lifting motion of all subjects. Markers were attached to body parts (see Fig. 1). Simultaneously, four pairs of surface electrodes were positioned bilaterally over longissimus dori (∼3 cm lateral to midline at the L1), external obliques (∼10 cm to midline above umbilicus and aligned with muscle fibers), rectus abdominis (∼3 cm to midline above umbilicus) and rectus femoris. The raw EMG signals were amplified and filtered at 30 Hz.

Subjects were instructed to hold no load, 30 N and 60 N in hands with a bar hanging weights. Subjects were expected to finish stoop lifting and squat lifting with 2 s pause at each posture shown in Fig. 3. For each subject, the experiment levels are 2 × 3. There are six repetitions for each level. Thus there are altogether 36 trials for each subject. For each three trials, subjects took one minute rest. One way ANOVA for repeated measure factors were used to study how BKAs and non-BKAs behave differently during lifting tasks. Three-way analysis of variance (ANOVA) for repeated measure factors were performed to study the effect of disability, lifting techniques (stoop lifting and squat lifting) and load (0 N, 30 N and 60 N) on EMG activities of extensor muscles and abdominal muscles. Besides, one-way ANOVA for repeated measure factors were used to study the effect of disability on EMG activities of the working muscles. Interactive effects between disability, lifting methods and loads were

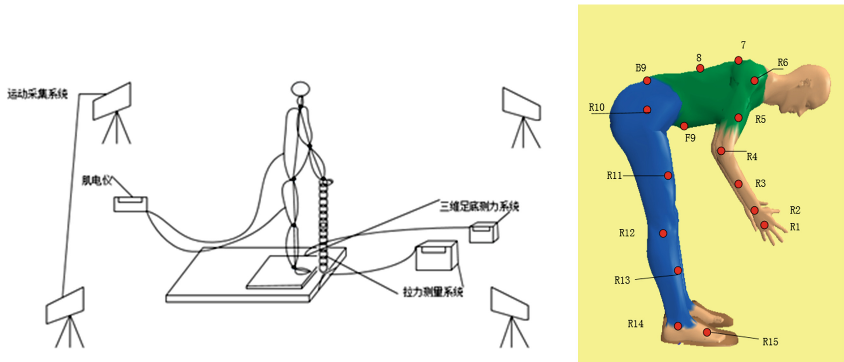

Fig. 1. Set of the experiment instrumented with 3D force plate, VICON motion capture and EMG electrode system. *Motion markers were attached to the body shown as dots on the body. Numbering both with number and letter means one pair on both sides of the body. R is on the right body part and L is for the left (unseen due to the cover from the right body part) plus number indicates the number of the right part of the body while numbering without letters*

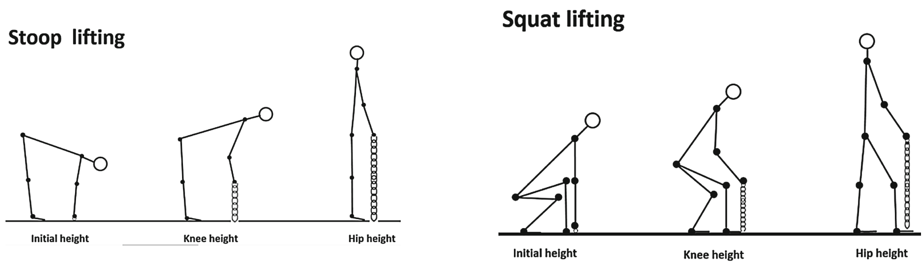

Fig. 2. Lifting postures for two lifting methods: (a) stoop lifting and (b) squat lifting. *Three stages of each lifting technique were paused for 2 s. In (a), subjects started to lift a weight with the arm and leg extended straight and marked as stage I; subjects reached stage II when the weight is at the same level with the knee; Stage III is when the subjects finish the lifting task with arm, lumbar and leg straight. In (b), during stage I, subjects started to lift a weight with the arm and lumbar extended straight and bent knees while the two feet were separated 30°from coronal plane; subjects reached stage II when the weight is at the same level with the knee; Stage III is when the subjects finish the lifting task with arm, lumbar and leg straight.*

evaluated to understand how disability, interacting with lifting methods and the loads in hands, affects muscle activities of extensor, abdominals and rectus femoris. Tukey's post hoc tests were performed to further reveal any significant trends ($p < 0.05$) (Fig. 2).

3 Results

After the experiment, the RMS of EMG was computed for each exertion by Matlab software. The percentage of maximum muscle EMG on extensors (the longissimus, iliocostalis), abdominals (external oblique, rectus abdominals) and rectus femora were compared with the normalized EMG (reflected as percentage of the maximum). The most apparent difference between BKAs and non-BKAs lies in longissimus as shown in Fig. 3. For BKAs, EMG activities in global extensor muscles (LGPT and ICPT) increased, though not significantly, in stoop lifting compared with those in squats lifting for the case with no loads in hands. EMG activity of extensor muscles significantly increased when the weight was added as 60 N and then for 30 N is not obvious. EMG muscles of rectus femoris changes most significantly in case of squat lifting with 60 N loads at hand although changes of rectus femoris is least in case of stoop lifting with no load at hand. Abdominal muscles, though relatively quiet in all tasks, demonstrated a significant change for two lifting methods and load (especially 0 N and 60 N). Disability has biggest effects on abdominal muscle activity across different lifting methods and loads. And the extensor muscles were affected least. Interactive effect between disability and lifting methods is more obvious than that between disability and loads. Tukey's tests revealed that the extensor muscle activity of BKAs increased more significantly from stoop lifting to squat lifting compared with non-BKAs. Activities of BKAs' abdominal muscles increased most significantly from stoop lifting to squats lifting compared to non-BKAs.

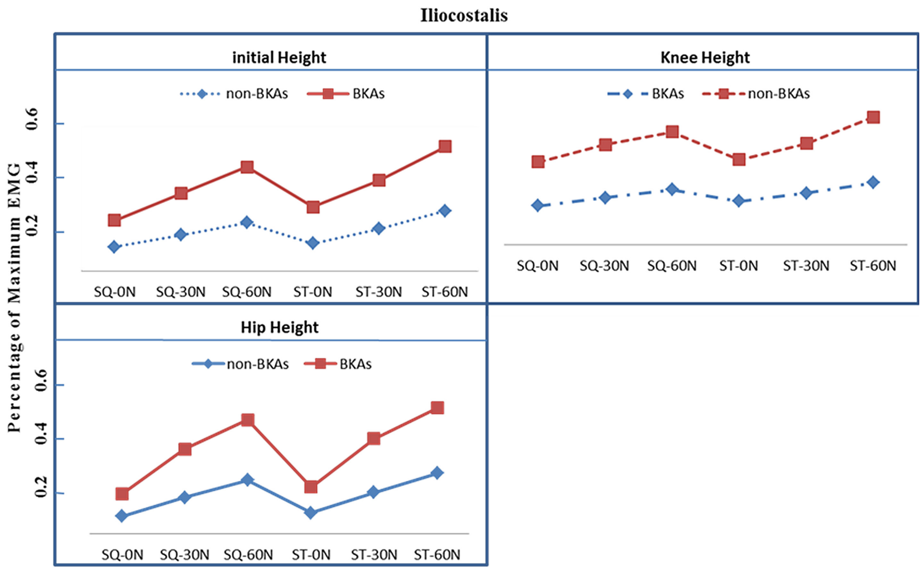

Fig. 3. Comparison of EMG of Longissimus between BKAs and non-BKAs

To determine whether the model calculating muscles forces specific for non-BKAs is still applicable to BKAs. Muscle forces of extensors, abdominal and femoral were predicted in AnyBody Modeling System™. "StandingModel" in the AMMRV1.3.1 was modified so that hand forces in "LeftArmDrivers.any" and "RightArmDrivers.any" were defined as 0 N, 30 N and 60 N (including the net weight of the bar). Body segment angles from electrogoniometers and sagittal photos were inputted to "Mannequin.any", and subsequently the inverse dynamic studies defined in "StandingModel. Main.any" were run using the infinite order polynomial optimization criterion. Similarly, "StandingModel" in the AMMRV1.3.1 was modified so that hand forces in "LeftFootDrivers.any" and "RightFootDrivers.any" were defined as the mean foot force recorded by 3D foot plate. Body segment angles from electrogoniometers and sagittal photos were inputted to "Mannequin.any", and subsequently the inverse dynamic studies defined in "StandingModel.Main.any" were run using the infinite order polynomial optimization criterion. The percentage of maximum muscle forces on the longissimus, iliocostalis, external oblique, rectus abdominals and rectus femoral were further calculated, and compared with the normalized EMG as shown in Fig. 4.

The percentage of maximum between EMG RMS and predicted muscle forces was not the same. In this study, the AnyBody Modeling System™ scaled the model linearly to fit the 50th percentile Asian population according to the subject's weight and height, so the maximum muscle force defined in the AnyBody Modeling System™ may not be the same as that of the individual subject. To solve the problem, the AnyBody Modeling System™ provides a mechanism to scale the model more accurately according to some external force measurements, individual segment length and weight.

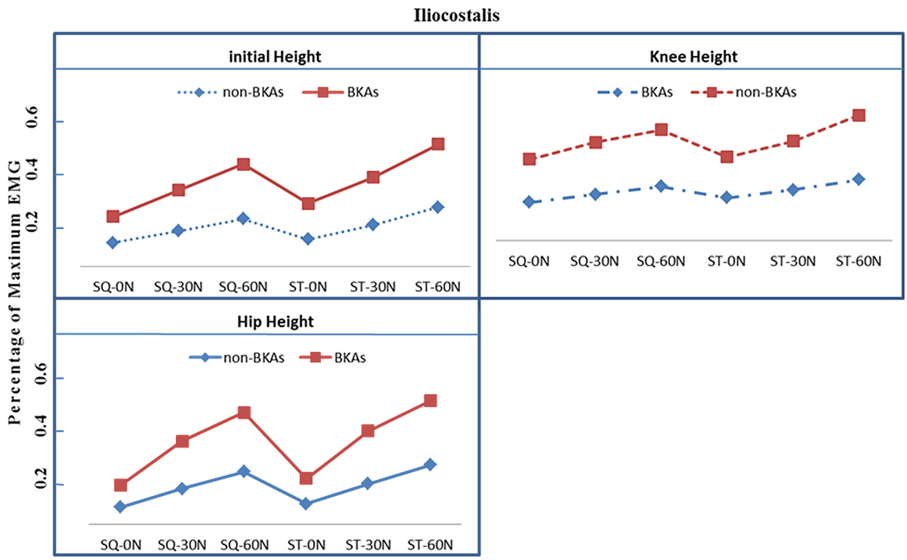

Fig. 4. Comparison of EMG of Longissimus between BKAs and non-BKAs

As expected, EMG and predicted forces of both muscles increased as hand loads increased in two lifting methods. The increasing trend of EMG was somewhat variable but the predicted muscle forces obey the similar trend. The Pearson's correlation coefficients between EMG and predicted muscle forces reached 0.7105. To improve the model predictability, further modification can be made to the model (Fig. 5).

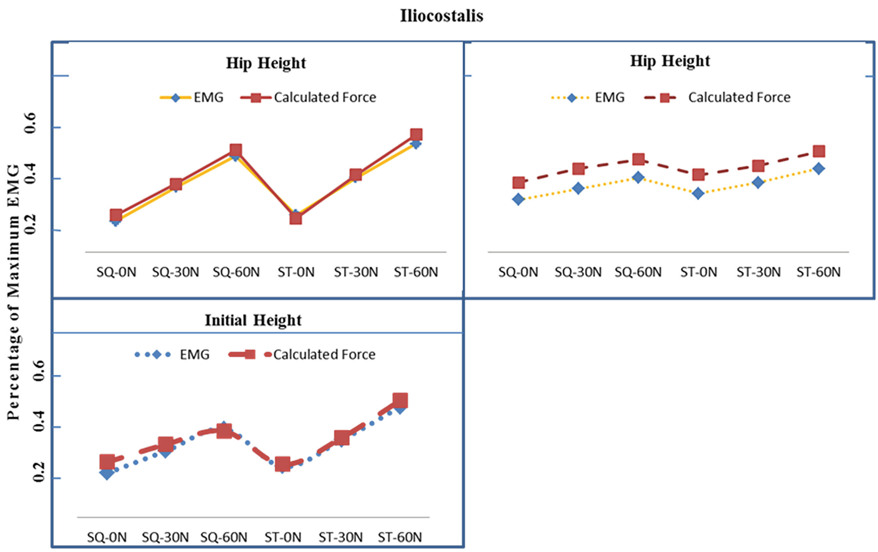

Fig. 5. Percentage of Maximum illiocostalis EMG RMS and Calculated Muscle of BKAs

4 Revised Modeling

There are two approaches currently for motion prediction: empirical statistical modeling and inverse kinematics or biomechanics. The first approach uses anthropometric data and motion patterns collected in the laboratory that are statistically analyzed to form a predictive regression model of posture with rule-based adjustments to accommodate the infinite motions possible. The second approach uses common inverse kinematics characterization to represent mathematically feasible postures. Inverse kinematics and optimization are used to assess the objective functions, such as joint limitations, physiology cost and thus generate the optimal posture/motion. In this paper, the mixture of the two approaches are applied to the algorithm of kinematic controller and dynamic controller to generate the comparative importance of each segment and optimize the posture functioned by the kinematic constraints and dynamic constraints (See Fig. 6).

Energy is the drive force of joint displacement while effort is a substitute to the changing posture from one point to another. Further optimization formulation is conducted to compute the factor of dynamic constraint for multi-FOD body segments.

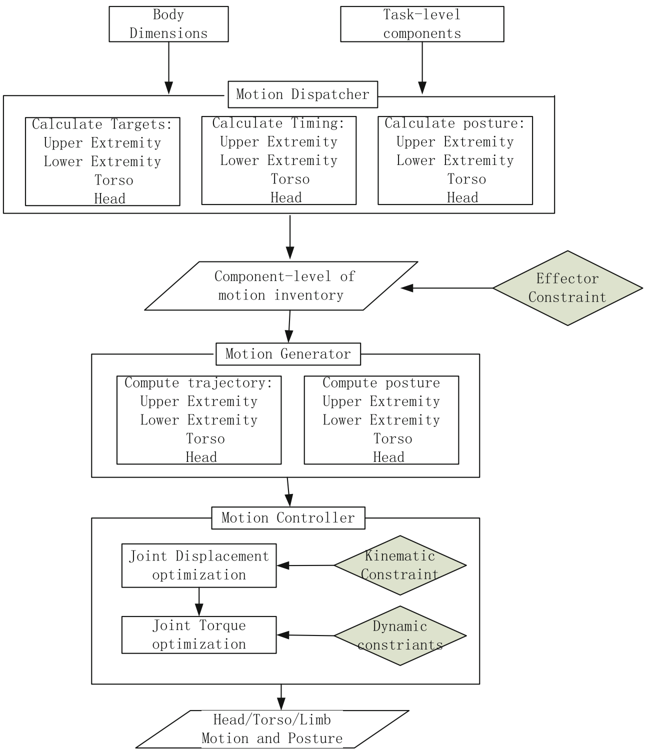

Fig. 6. Task Modeling Process

Joint Displacement Profile:

$$F\left(q'\right) = \sum\nolimits_{i=1}^{n} w_i (q_i' - q_i)^2 \tag{1}$$

To minimize:

$$F(\tau) = \sum\nolimits_{i=1}^{n} w_i' |\tau_i| \tag{2}$$

St: $F(q') \in F(q)$

$$\tau_i^L \le \tau_i \le \tau_i^U$$

w_i' is the deviation caused by the physical constraints relative importance of each joint in the motion.τ_i can be calculated as the following:

$$\tau_i = M_{ik}(q)\,\ddot{q} + \sum J(q_i)^+ m_{ik} g + \sum J(q_k)^+ F_k \tag{3}$$

i = 1, 2, …, n.

$$M_{ik}(q) = \sum\nolimits_{j=\max(i,k)}^{n} \mathrm{Tr}\left\{ \frac{\partial T_j(q)}{\partial q_k} I_j \left[\frac{\partial T_j(q)}{\partial q_i} \right] \right\} \tag{4}$$

i, k, j = 1, 2, , ... , 3

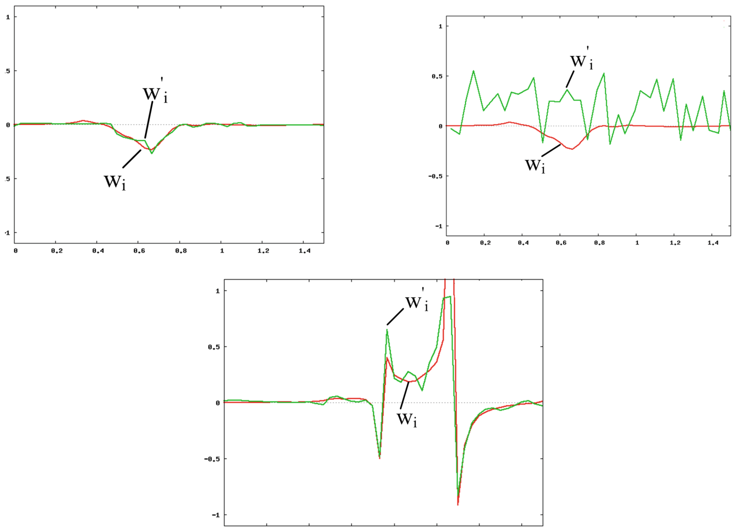

Fig. 7. The DOF weights of knee(a), L5/S1, and L3 over time. The red curve represents the value of wi and the green represents the value of w_i'.

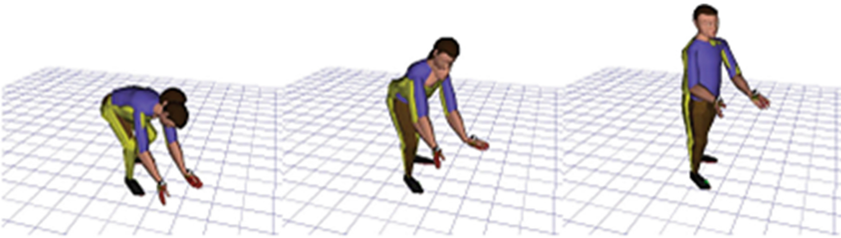

Fig. 8. Comparison of captured (yellow shirt) and modeled (blue shirt) task postures of the subject (Color figure online).

m_{jk} is the mass of link (i,k), $M_{ik}(q)$ is the mass inertia of link (i,k). F_k is the external force on the joint k. Joint i and k are the two joints on each side of the link (i,k).$\tau = \tau(t)$, $F_k = F_k(t)$, and $q = q(t)$.

The EMG trends and difference from calculated force in Sect. 3 were used to train neural network to get a satisfactory w_i and w_i'. The trained results of (w_i, w_i') are shown in Fig. 7.

Task simulation on one subject was used to explain the validity of the revised model. Manipulated by the weights at each corresponding time point, the model put out the optimization angles of 6 joints. The calculated result was put into Jack environment and a manikin was created, which was compared to another manikin only created by motion capture data. The matching results were shown in Fig. 8. As Fig. 8 shows, the yellow shirt is almost overlapped with blue shirt. The most obvious mismatching between the yellow shirt and blue shirt lies in the posture of squatting. As for the other postures, the mismatch is not observed obviously. Disparity occurs when the physical constrained part is required to exert great effort to implement the motion/posture. The variation might be caused by the weight obtained from neural network training from small number of subjects. Further study can be conducted calculating the weights with more subjects.

5 Conclusion

In an attempt to search for the safer lifting method, the study aimed to investigate the relative muscle activity difference between BKAs and non-BKAs. It is notably found that disability has greatest effects when lifting bigger weights and BKAs performed quite differently in term of muscle activities in case of squat lifting. Abdominal muscles reflected the difference most obviously.

Modeling examples shows that for lifting tasks simulated by the AnyBody Modeling System™, the infinite order polynomial (min/max) cannot predict muscle forces correlated well with the EMG due to disability. The framework proposed here reproduces disabilities into three levels: effecter, kinematic and physical, and optimize the position and force of the physical handicapped through motion controller and modified the calculated model. The results simulated in JACK shows good fidelity. The unsatisfactory part of the results lies in the validity of the weights and simplified kinematic model with limited FOD for each joint. The future work can focus on the enhancement of our weight based constraint model by enlarging more samples and set up a kinematic skeleton based on careful observation of the real motion which definitely require more FODs for each body link and joint.

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