Glenohumeral contact forces during five activities of daily living

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ABSTRACT - Five activities of daily living were studied in order to determine the glenohumeral contact forces during common high-load activities. These activities included standing up from and sitting down into a chair using the arms, walking with a cane, lifting a 5 kg box using both hands and lifting a 10 kg suitcase. The movements and hand loads of six healthy older subjects were recorded with a motion analysis system from which trunk and arm joint angles were calculated. This information was input into a biomechanical model developed by Peterson et al. (1994). The resulting contact forces ranged from 0.3 to 6.9 times bodyweight with typical values ranging from two to three times bodyweight. The highest average contact force was associated with lifting a suitcase. Since higher loads may be carried this is likely the most important task for high glenohumeral contact forces. The contact force values should be used with caution as there is some concern about the validity of the model under high loads and in the particular positions examined. This study was performed in order to derive a suitable load for testing shoulder prostheses.

INTRODUCTION

Total shoulder prostheses fail primarily by instability and glenoid loosening (Wirth & Rockwood, 1996). Both of these complications are related to loading on the implant as well as other factors. To test for these failures it is necessary to know the maximum and repetitive loads to which the glenoid is subjected. Until recently the prime source of contact forces at the shoulder has been a study by Poppen and Walker (1978) which predicted a maximum force of 0.9 times bodyweight (BW) for pure abduction in the scapular plane. This was based on a simple two-dimensional analysis and only three specimens but is widely reported because few alternatives existed. More recently several three-dimensional models have been developed. The only three-dimensional tasks that have been studied however are push-ups, chin-ups and press-ups (Runciman, 1993) as well as wheelchair propulsion (Van der Helm & Veeger, 1996). Unfortunately, none of these would commonly be performed by a person with a shoulder prosthesis.

Three-Dimensional Biomechanical Models

Three major musculoskeletal models of the shoulder have been developed (Karlsson & Peterson, 1992; Runciman, 1993; Van der Helm, 1994). The models differ with respect to the number of muscles and muscle elements, the muscle attachment points, the cross-sectional areas of the muscles, the scapulohumeral rhythm, anthropometric measurement inputs, static versus dynamic capabilities and the stability constraint. In all cases the glenohumeral joint is treated as a spherical ball-and-socket joint with a fixed center of rotation. Validation has mostly been limited to qualitative comparisons with EMG measurements. Makhsous (1996) also found reasonable agreement in the Swedish model for intramuscular pressure and maximum strength.

Runciman (1993) studied five male subjects performing three activities. He calculated peak glenohumeral contact forces for push-ups to average approximately seven times BW when a joint stability constraint was included. There was a wide variation among individuals. The force was directed almost normal to the glenoid fossa. For both chin-ups and press-ups (similar to rising from a chair but using the arms only), the peak contact force averaged over four times BW.

Van der Helm and Veeger (1996) report maximum joint contact forces of 1900 N when the hand is at its

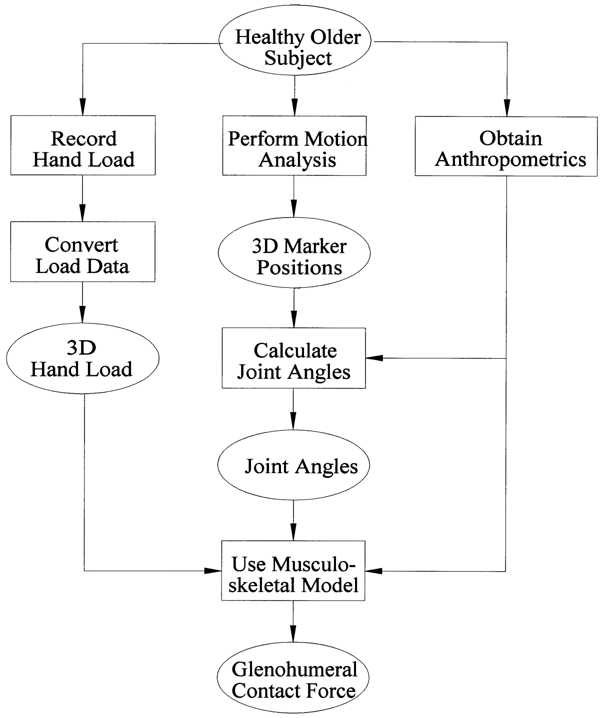

Figure 1- *Analysis flowchart*

Table 1- *Average model input data*

Input Variable	Standing	Sitting	Cane	Box	Suitcase
Hand Load (N)	134	114	130	-25	-128
Load Vector (N)	(1,22,131)	(2,24,96)	(4,10,128)	(0,0,-25)	(0,0,-128)
Wrist-Load Distance (mm)	58	58	64	113	88
Clavicle:Pro/Retr Angle	-18	-18	-23	-28	-24
Clavicle: Elev/Depr. Angle	21	21	9	20	10
Clavicle: Axial Rotn	15	15	8	-6	-3
Scapula:Pro/Retr. Angle	45	45	38	42	44
Scapula:Lat. Rotn Angle	7	7	7	15	2
Scapula: Tipping	21	21	20	12	17
Trunk Flex/Ext. Angle	-31	-33	-10	-1	-34
Plane Angle	-35	-43	-14	60	35
Elevation Angle	34	30	11	70	25
Rotation Angle	72	66	45	-43	-26
Elbow Angle	84	94	55	62	34
Pro/Supination Angle	54	49	23	-34	11

highest point on a wheelchair handrim while applying 40% of the subject's maximum voluntary moment on a static wheelchair ergometer.

Motion Studies

Using the arms to aid in rising from a chair has been shown to reduce forces at the knee by at least 18% (Ellis et al., 1984). Okuno et al. (1995) found that using a cane reduced bone-on-bone forces at the knee by 20 to 30% during the peak loads of the stance phase. Thus, while loads at the knee are typically high, a significant portion of this load may be transferred to the shoulder via a cane or armrests to relieve the knees.

Upper arm angles have been studied by several groups, however the majority did not study functional tasks, none studied the tasks defined here and no standard measurement protocol has been developed. Almost all of the quantitative studies have been conducted in the last six years.

The shoulder is typically considered to be exposed to only relatively small (less than bodyweight) loads, based on analyses of pure abduction with a light load. It was the aim of this research to study common tasks that would expose the shoulder to higher loads with the purpose of using this information to test shoulder prosthesis components.

MATERIALS & METHOD

Five tasks were performed by each subject:
1) standing up from a chair using the arms,
2) sitting down into a chair using the arms,
3) walking with a cane,
4) lifting a 5.0 kg box from the floor to shoulder height using both hands, and
5) lifting a 10.0 kg suitcase.

These tasks were chosen to reflect common activities which would lead to high shoulder forces. Sitting and standing were divided into two tasks as a short pause is insufficient to separate the tasks (Packer et al., 1993). Each task was repeated five times with minimal rest between tasks. The vertical and two shear forces exerted on the chair arm were measured by a transducer built into the hand support while forces on the cane were measured by an AMTI force platform. The chair was a standard chair with a seat height of 44 cm and an arm height of 66 cm. The cane was adjustable in 25 mm intervals and was set to the height most natural for the subject. The box was 42 cm by 15 cm by 16 cm high. The suitcase was 65 cm by 18.5 cm by 51 cm high with an additional 5.5 cm for the handle.

Six healthy subjects, three male, three female, were recruited for the study. They were required to be 50 years of age or older with no history of shoulder problems. Ages ranged from 51 to 64 (mean, 55), weight from 52 to 89 kg (mean, 73 kg) and height from 152 to 187 cm (mean, 168 cm). All subjects except one were right-hand dominant.

Figure 1 outlines the approach used to derive the glenohumeral contact forces. Infrared emitting diodes (IREDs) were affixed with double-sided tape to the skin at five specified locations on the lateral side of the right arm (Anglin, 1993; Romilly et al., 1994). The shoulder joint center was based on the definition of Wang (1996), namely 37 mm inferior, 14 mm lateral and 8 mm anterior to the acromion. A plastic 8 cm square extension was affixed perpendicular to the trunk at approximately the first three thoracic vertebrae to detect trunk motion. Two markers were aligned on the trunk extension in the vertical direction; the third was out of plane in the posterior direction. The positions of the markers were recorded with the OPTOTRAK motion analysis system from Northern Digital. The distance from the wrist to the load was measured for each task.

Since the scapula and clavicle move under the skin, surface markers are inappropriate and only static positions could be recorded. Following the tasks, the positions were recorded using the OPTOTRAK Stylus. This device, with six locating markers, was pointed in sequence at each of the normal markers plus four additional palpated locations: point of the acromion, inferior angle of the scapula, root of the spine of the scapula and the sternoclavicular joint. Three additional locations were palpated but ultimately not used. None of these are needed for the Swedish model which uses a predefined scapulohumeral rhythm. Due to the sequential recording it would have been too fatiguing to either hold a load or one's bodyweight thus only positions at the beginning and end of each task were recorded.

Given the three-dimensional positions of the markers (Costigan et al., 1992), the joint angles of the arm were calculated using a direct geometrical analysis (Anglin, 1993; Romilly et al., 1994). Several additions were made from the previously described analysis. Trunk movement was allowed and measured. Corrections were also made to the elbow flexion and upper arm rotations to accommodate the pro/supination axis passing through the distal ulna rather than midway between the styloid processes. Joint angle accuracy was estimated to be within 6 degrees.

Except for lifting the suitcase, the loading was assumed to be static due to the slow accelerations. To calculate the dynamic suitcase load, the accelerations were calculated from the hand marker and added to the gravitational acceleration. The joint angle information combined with the loading at the hand and the anthropometrics of the subjects were input into the biomechanical shoulder model (Table 1) developed by Karlsson & Peterson (1992) and Peterson et al. (1994). When an unfeasibility was encountered at high loads, the muscle parameters were multiplied by up to two times, which was sufficient in all cases except the sitting task for subject 2. A 25 degree conical stability constraint was used to maintain the contact force within the glenoid fossa.

RESULTS

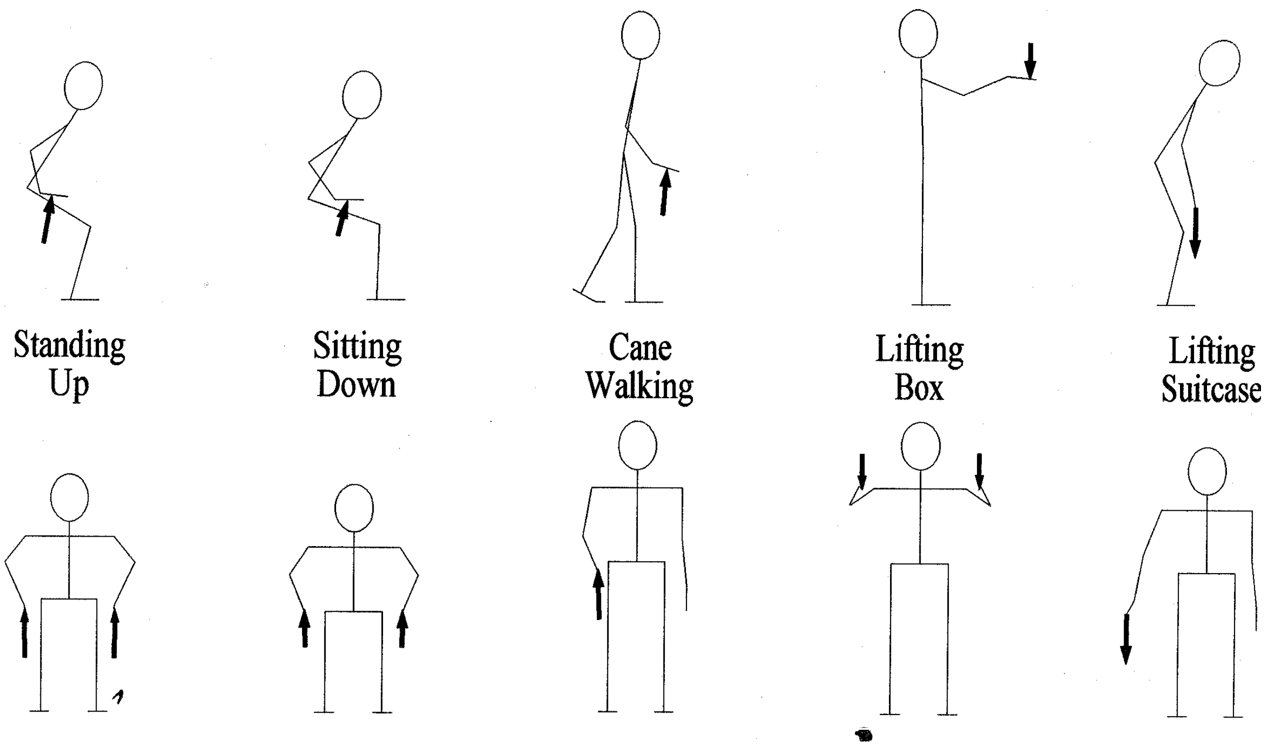

Figure 2 - *Stick figures of average joint angles and hand loads.*

Table 2 - *Glenohumeral contact forces as multiple of bodyweight*

	Standing	**Sitting**	**Cane**	**Box**	**Suitcase**
Average (All Trial & Subjects)	1.8	2.0	1.7	1.8	2.4
Range (All Trials & Subjects)	0.5 - 4.3	0.3 - 6.9	0.4 - 3.2	1.5 - 2.3	1.3 - 4.3
Avg. Max. (Across Trials)	2.7	2.9	2.0	1.9	2.7
Max. Avg. (Across Subjects)	3.1	5.1	2.8	2.2	3.5

Table 1 presents the average model input data based on the results of the motion analysis. Limb lengths and masses were also changed to suit each individual. The scapular and clavicular data was not required for the Swedish model. Figure 2 represents the average data graphically. To avoid the effect of outliers, the highest and lowest values were removed to calculate the mean. The angles and loads correspond to the point of peak hand load. One position and load were analyzed for each of the five tasks for all six subjects and all five trials. The first coordinate of the load vector is positive to the right, the second positive forward and the third positive up. Angular data follows definition 1b in Van der Helm & Pronk (1995). The arm joint angles are relative to the thorax and represent sequential Euler rotations. Trunk movement other than flexion was only significant for lifting the suitcase. Since bending to the side reduces the horizontal moment arm of the load, the contact force in Table 2 is overestimated. The plane-of-elevation angle is relative to the frontal plane. This angle has been given many other names (azimuth, horizontal flexion/extension, pole angle) and it is hoped that a standard name will come into usage. Elevation is rotation out from the thorax or humeral rest position. Rotation is axial rotation of the humerus, which is only equal to anatomical internal/external rotation when the plane of elevation is zero. Elbow flexion is defined to be zero in full extension. Pro/supination is positive in the pronation direction. Wrist flexion and radial/ulnar deviation were also calculated but are not included as they do not affect the shoulder calculations.

Table 2 summarizes the contact forces normalized for bodyweight. Figure 3 divides the contact force results by subject and task. The average maximum in Table 2 refers to selecting the trial with the highest contact force for each individual (i.e. the maximum of each line in Figure 3) and averaging them. This likely gives the best representation of the highest expected contact force. The maximum average refers to selecting the subject with the highest average contact force. This best represents the outer limit that would be expected across a population. A sensitivity analy-

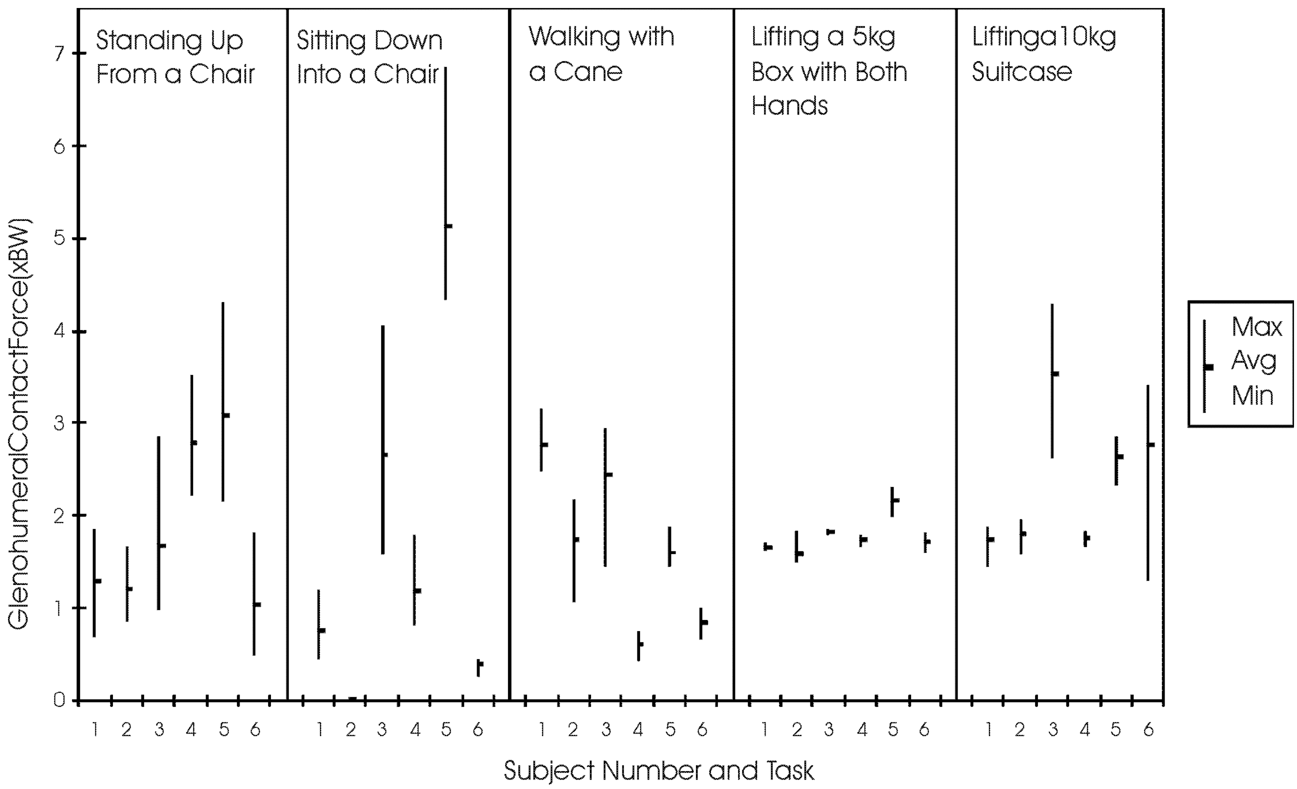

Figure 3 - *Glenohumeral contact forces by subject and task.*

sis was conducted by taking the average data for each task and perturbing each angle except pro/supination (which had a negligible effect) by five degrees in both directions. All combinations (35) of the angles were input into the model. The resulting minimum and maximum contact forces for the five tasks, following the order in Table 2, were 0.6 to 2.3, 0.7 to 3.7, 0.9 to 2.8, 1.7 to 2.0 and 1.1 to 3.5 times BW. This revealed that the model was sensitive when a muscle reached its defined maximum, which occurred for some subjects in all tasks except lifting the box. For all tasks except lifting the box the angle of the contact force away from the normal to the glenoid fossa was 25 degrees, equal to the stability constraint.
The box and suitcase tasks were analyzed for different loads. The relationship between load and contact force was linear for both so the contact force for other loads can be easily calculated assuming the same body position. For lifting a box the linear regression was:

- CF = -32.7L+479 (R^2=0.998)

where CF is contact force in Newtons and L is the negative load in Newtons (half the box load since both hands are used); for lifting a suitcase the regression was:

- CF = -11.5L+171 (R^2=0.999).

DISCUSSION

The resulting maximum glenohumeral contact forces appear to be between two and three times bodyweight with some higher values. Lifting a suitcase has the highest average contact force, although the inclusion of bending to the side would lower this value. Since subjects certainly might choose to lift more than 10 kg and at a greater elevation, however, this could lead to even higher loads. Falling onto the arms would lead to higher loads than found here but may lead to overdesign of a prosthesis if used as a requirement. Although only six subjects were studied they represent a wide variety of heights and weights and both sexes and thus adequately describe the variation that could be expected in the able population.
As has been commonly found in motion studies, intersubject variability is high. Whereas the cane and shelf height were adjusted for each individual, the chair height remained the same requiring different kinematics for each subject. This was intentional since the majority of chairs are not adjustable. Intrasubject variability is also high for the chair tasks and for some subjects in the cane and suitcase tasks. The box lifting task is very repeatable. This within-subject variability may be as much due to sensitivity of the model at high loads as to the variations between trials.
The two most difficult problems in modeling the shoulder are individual anthropometrics and the scapulohumeral rhythm, as has been discussed by each of the model developers. It would be impractical to measure muscle parameters on the living subjects, yet average values are suspect due to the wide variability that exists (Van der Helm et al., 1992). Similarly, while an average scapulohumeral rhythm has been questioned, measuring clavicular and scapular points in static positions to represent dynamic motions is also questionable.
The primary concern with the shoulder model is the validity of the results. In general, the validation studies by Makhsous (1996) are encouraging. However, the model has not been validated at higher forces and in the positions examined. The sensitivity at higher loads suggests that the resulting values should be used with caution. In order to avoid unfeasibilities in the model, the muscle parameters were uniformly scaled up, by up to two times, until a solution was achieved. This was done for some subjects in all tasks except lifting the box. This appeared to give a more reasonable result than extrapolating contact forces calculated at lower load levels. In the case of the chair tasks, this resulted in an unreasonably high triceps cross-sectional area (56 cm^2 for all three parts), particularly given that the subjects were of an older age. Unfeasibilities occurred when one or more muscles were at the defined maximum load that they could carry, yet presumably there were not sufficient alternative muscles to supply the extra force. The triceps prime purpose is elbow extension, but in crossing the shoulder joint it contributes to the joint contact force. The lack of a developed elbow in the Swedish model may have overestimated the force in the triceps and hence the contact force at the shoulder. Preliminary use of the Dutch model (Van der Helm, 1994) suggested much lower contact forces. Use of the Dutch model as well as a comparison of the two models will be pursued in the future.
A contact force of two to three times bodyweight is comparable to that for wheelchair propulsion (Van der Helm & Veeger, 1996) and less than that for chin-ups, press-ups and push-ups (Runciman, 1993) as would be expected. It is greater than for pure abduction and less than is typical at the knee and hip. Thus the findings of this study, although perhaps overestimated, are nevertheless in rough agreement with other studies.

CONCLUSIONS

Five common activities of daily living were studied culminating in the calculation of the contact force on the glenoid. These tasks, chosen to result in high shoulder forces, led to contact forces of between 0.3 and 6.9 times bodyweight, with typical maximum values between two and three times bodyweight. Three times bodyweight translates to 1500 N to 2600 N for the subjects in this study. These values suggest loads that could be applied in either prosthetic testing or finite element analyses. Excessive loading, either as a one-time large force or at a lower repetitive level, may contribute to glenoid loosening. Mechanical testing of prosthesis components for subluxation and glenoid loosening is planned for the near future. The detailed analysis of arm tasks, as outlined in this paper, will support a load selection for standardized testing of components. Further work with both the Swedish and Dutch models will investigate the differences between the results of the two models.

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