Glenohumeral contact forces in reversed anatomy shoulder replacement

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A B S T R A C T

A major requirement to design an implant is to develop our understanding of the applied internal forces during everyday activities. In the absence of any basic apparatus for measuring forces directly, it is essential to rely on modelling. The major aim of this study was therefore to understand the biomechanical function of subjects with the reversed anatomy Bayley–Walker prosthesis, using an inverse dynamic shoulder model. In this context, the muscle and joint forces of 12 Bayley–Walker subjects were compared to those of 12 normal subjects during 12 activities of daily living.

Maximum glenohumeral contact forces for normal and Bayley–Walker subjects were found to be 77% (±15) and 137% (±21) body weight for lifting a 2 kg shopping bag, and the least forces 29% (±4) and 67% (±8) body weight for reaching to opposite axilla, respectively. For normal subjects, middle deltoid, supraspinatus and infraspinatus were found to be the most active muscles across the subjects and tasks. On the other hand, for implanted subjects with a lack of rotator cuff muscles, the middle deltoid and coracobrachialis muscles were found to be the most active. The biomechanical model can therefore be used in order to gain knowledge about the pathology as well as possible post surgical rehab for subjects with reversed shoulder replacement.

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1. Introduction

Due to the steady growth in the ageing population, the number of people with joint diseases is growing and therefore the topic of joint replacement is becoming more important. In order to design a replacement extensive knowledge about the joints contact forces is required. The difficulty is that these forces cannot be measured directly; consequently we need to rely on modelling. The internal forces for natural anatomy have been previously analysed using biomechanical models with different complexities (Charlton and Johnson, 2006; Dickerson et al., 2007; Favre et al., 2005; Karlsson and Peterson, 1992; Runciman and Nicol, 1994; Van der Helm, 1994). These are inverse dynamic models predicting muscle and joint forces from kinematic data. Anglin et al. (2000) used the Swedish Shoulder Model (Karlsson and Peterson, 1992) and reported glenohumeral force (GHF) of up to 240% of the body weight for normal anatomy shoulder during everyday task where the external load is applied to the arm.

Rotator cuff (RC) tear is a common cause of pain and function reduction among adults (Sher et al., 1995). One popular approach for treatment of RC tear is the reverse anatomy shoulder replacement (Katz et al., 2007). Despite the long term use of reverse anatomy shoulder prostheses, only a few published studies are available that have analysed their biomechanics (Kontaxis and Johnson, 2009; Masjedi and Johnson, accepted for publication; Terrier et al., 2008; Van der Helm, 1998). A limitation of these studies is that the internal forces have only been calculated for standardised movements such as flexion and abduction, rather than more meaningful activities of daily living (ADL). The possible difference between using ADL and standardised activity has been pointed out previously by Bergmann et al. (2007) who designed an instrumented implant (anatomical prosthesis) capable of measuring the six components of GH forces and moments. The data from the instrumented anatomical implant were taken and reported for 69 year old, 100 kg male subject and the GHF of below 100% of the body weight was reported for most of the ADL. They compared their data with the Swedish Shoulder Model (Karlsson and Peterson, 1992) and reported that although a good agreement was observed for standardised activity such as abduction, with more complex activities the models over-estimated the forces.

The main aim of this study was to analyse and understand the biomechanical function of subjects with the reversed anatomy Bayley–Walker (B–W) prosthesis during everyday activities, using an inverse dynamic shoulder model (Charlton and Johnson, 2006). Additionally use of such a model to understand the shoulder pathologies and to help the clinician for pre and post surgical decision making was evaluated. The applied model (Newcastle shoulder model—NSM) is an inverse dynamic model of the shoulder complex with representation of 31 shoulder muscles.
(Johnson et al., 1996), while the skeletal geometry is taken from the Visible Human Project (Spitzer and Whitlock, 1998). In this model, a constraint was used to satisfy the stability criterion of the GH joint; the shape of glenoid was approximated as an ellipse and the GH joint forces were constrained to pass from this area. The model has previously been modified to include the B–W prostheses (Masjedi and Johnson, accepted for publication). The modifications included changes in geometrical parameters as well as modification of stability region of the GH joint (a hemispherical surface is used for the B–W implant instead of ellipse). The centre of rotation for subjects with the B–W prosthesis is similar to that of normal anatomy and the middle deltoid moment arm therefore remains almost unaltered (Masjedi and Johnson, accepted for publication).

2. Methodology

The kinematics of 12 subjects with the B–W prosthesis (2 M/10 F mean age 71.3 ± 11 years) have been analysed previously and compared with the kinematics of 12 subjects with no shoulder pathology (10 M/2 F, mean age of 43 ± 15.8 years) (Masjedi and Johnson, 2008). In this study a motion capture system (Vicon) was used to record the position of 10 markers attached on palpable bony landmarks of natural anatomy and the middle deltoid moment arm therefore was therefore simulated by setting the maximum potential force generation of RC attachment sites as well as the muscle lengths at each instant. No information was available regarding the B–W subjects’ RC deficiency, it was therefore assumed that all subjects had full RC tear. In the NSM, the maximum potential force of a muscle (element) depends on its physiological cross sectional area (PCSA). The tear was therefore simulated by setting the maximum potential force generation of RC muscles to zero. By employing the dynamic and muscle parameters data as an input to the model, an optimisation algorithm based on the minimisation of the sum of squared muscle stresses was used to predict the muscle and joint forces for both B–W and normal subjects.

Since the scapula and clavicle regression equations (Barnett et al., 1999; Marchese, 2000) used in the NSM are not valid for the tasks performed behind the coronal plane, Task 3 (washing lower back) was omitted from this study. The outcome from GHF and its line of application will be presented for both B–W and normal subjects in the distal co-ordinate frames. This will be followed by a presentation of the estimated forces in the muscles (that cross the GH joint) and their activation level. Muscle activation was defined by Charlton and Johnson (2006) as predicted muscle force divided by maximum possible force in that muscle. Finally, similar to Praagman et al. (2000) who found a linear relationship between the GHF and the net moment predicted by the Delft Shoulder Model (DSM) (Van der Helm (1991)), a linear regression equation was created to take the net moment data as an input and predict the GHF for both normal and the B–W subjects (B–W–S). Net joint moment calculated by the root of the sum of squared components of moments around the three axes.

3. Results

3.1. Glenohumeral contact force and its point of application

The mean GHF results for the normal and B–W–S for each activity are shown in Fig. 1 with one range of standard deviation. These results show that in all activities the GHF, as well as the inter-subjects variation for B–W–S, were greater than those of normal subjects. For the normal subjects the highest forces were observed in Task 8 and the least total forces were observed in Task 2. The average GHF magnitude across all subjects and across all tasks had the maximum value of 296 N. For B–W–S the highest force was also observed in Task 8. The average contact force magnitude across all subjects and across all tasks reached the maximum value of 400 N which was almost 100 N greater than that of normal subjects. The greatest difference for the mean maximum GHF between the two groups was observed for Task 8 followed by Task 6, with a value of 400 and 328 N, respectively. Furthermore for all activities the GHF point of application was within the constrained boundaries (hemisphere for B–W subjects and ellipse for normal subjects) subjects in both groups.

The overall mean (averaged across subjects and then task cycle) GHF for each activity across subjects of both groups are shown in Fig. 2. The highest forces were observed for Task 8 while the least forces were seen in Tasks 2 and Task 7 for normal and B–W–S, respectively. The overall mean GHF (average across subjects and activities) was found to be 40% (±11%) of the body weight for normal subjects and 74% (±21%) of the body weight for B–W–S.

3.2. Muscle forces

**Normal subjects:** The shoulder muscle forces for each activity averaged across the subjects are obtained and the maxima are shown in Table 2. The highest muscle force was observed for Task 8 in the supraspinatus muscle with a value of 146 N. In general, the largest muscle forces across all activities were observed in the

<table>
<thead>
<tr>
<th>Task protocol</th>
<th>Activities of daily living</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Mug to mouth (drinking)</td>
<td>7. Pouring from kettle standing (5 N)</td>
</tr>
<tr>
<td>2. Reach to opposite axilla</td>
<td>8. Lift shopping bag (2 kg)</td>
</tr>
<tr>
<td>3. Wash lower back</td>
<td>9. Lift tray (0.5 kg) use both hands</td>
</tr>
<tr>
<td>4. Brush opposite side of head</td>
<td>10. Sitting position lift to shoulder height (0.5 kg)</td>
</tr>
<tr>
<td>5. Answer telephone</td>
<td>11. Reach as far as you can</td>
</tr>
<tr>
<td>6. Pouring from kettle sitting (5 N)</td>
<td>12. Sitting position lift to head height (0.5 kg)</td>
</tr>
</tbody>
</table>
supraspinatus followed by the middle deltoid and infraspinatus and the smallest forces were observed for subscapularis muscles.

B–W subjects: The deltoid force has increased substantially for all activities relative to those of the normal subjects (Table 2). Middle deltoid has reached its capacity (230 N) for all B–W–S while performing Task 8. Furthermore, the anterior portion of the deltoid frequently reached its capacity; 190 N for Task 8, which increased substantially in comparison to mean maximum of 38 N.

Fig. 1. (A) Comparison of the GHF magnitude and its variation (± standard deviation) across subjects for Tasks 1–7. (B) Comparison of the GHF magnitude and its variation (± standard deviation) across subjects for Tasks 8–12. Bottom left shows the averaged GHF across the subjects and then activities.
for normal subjects. The predicted force in the posterior deltoid from the mean maximum of 50 N for normal subjects increased to 200 N for B–W–S during Task 3 (brushing the hair). The forces in coracobraclialis muscle were also found to be generally large and are shown in Table 2. For full details regarding the trends of each muscle force during the ADL, see Ph.D. thesis by Masjedi (2009).
The normal group.
were found to be less active in the B–W–S compared to those of
capulo-thoracic muscles such as trapezius and serratus anterior
the middle deltoid was found to be the most active muscle

Fig. 2. Overall mean glenohumeral contact force, averaged across normal and
B–W subjects. (Error bars represent the standard deviations.)

Table 2
Mean maximum glenohumeral muscle forces during activities of daily living
averaged across normal and B–W subjects.

<table>
<thead>
<tr>
<th>Tasks</th>
<th>Muscle/strings</th>
<th>Normal subjects maximum muscle forces (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Deltoid anterior 1</td>
<td>8 7 21 8 18 3 21 7 13 13 22</td>
</tr>
<tr>
<td></td>
<td>Deltoid anterior 2</td>
<td>16 13 26 15 29 4 38 10 24 20 31</td>
</tr>
<tr>
<td></td>
<td>Deltoid middle</td>
<td>50 50 84 63 94 57 107 66 73 87 84</td>
</tr>
<tr>
<td></td>
<td>Deltoid posterior 1</td>
<td>0 2 7 1 0 7 20 10 2 32 9</td>
</tr>
<tr>
<td></td>
<td>Deltoid posterior 2</td>
<td>5 8 23 13 10 30 53 20 12 53 29</td>
</tr>
<tr>
<td></td>
<td>Supraspinatus</td>
<td>74 56 75 83 100 68 148 65 100 107 98</td>
</tr>
<tr>
<td></td>
<td>Infraspinatus 1</td>
<td>18 20 27 23 40 25 56 26 27 33 33</td>
</tr>
<tr>
<td></td>
<td>Infraspinatus 2</td>
<td>31 40 52 38 75 42 91 44 47 57 61</td>
</tr>
<tr>
<td></td>
<td>Infraspinatus 3</td>
<td>36 48 66 41 94 39 101 45 52 66 75</td>
</tr>
<tr>
<td></td>
<td>Subscapularis 1</td>
<td>119 136 199 79 145 48 165 77 107 98 136</td>
</tr>
<tr>
<td></td>
<td>Subscapularis 2</td>
<td>0 0 1 0 0 0 4 12 7 15 1</td>
</tr>
<tr>
<td></td>
<td>Subscapularis 3</td>
<td>18 20 27 12 17 2 19 2 3 10 15</td>
</tr>
<tr>
<td></td>
<td>Teres minor</td>
<td>1 4 6 1 17 2 19 2 3 10 15</td>
</tr>
<tr>
<td></td>
<td>T.maj</td>
<td>0 0 1 0 0 1 2 3 1 0 0</td>
</tr>
<tr>
<td></td>
<td>Coracobrachialis 1</td>
<td>2 5 11 3 5 2 6 4 4 2 4</td>
</tr>
<tr>
<td></td>
<td>Coracobrachialis 2</td>
<td>2 5 12 3 5 2 6 4 4 3 5</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Tasks</th>
<th>Bayley– Walker subjects maximum muscle forces (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Deltoid anterior 1</td>
</tr>
<tr>
<td></td>
<td>Deltoid anterior 2</td>
</tr>
<tr>
<td></td>
<td>Deltoid middle</td>
</tr>
<tr>
<td></td>
<td>Deltoid posterior 1</td>
</tr>
<tr>
<td></td>
<td>Deltoid posterior 2</td>
</tr>
<tr>
<td></td>
<td>T.maj</td>
</tr>
<tr>
<td></td>
<td>Coracobrachialis 1</td>
</tr>
<tr>
<td></td>
<td>Coracobrachialis 2</td>
</tr>
</tbody>
</table>

The overall muscle activations averaged across subjects in both
groups and then across tasks, are shown in Fig. 3. In the B–W–S
the middle deltoid was found to be the most active muscle
followed by the coracobrachialis muscle. Furthermore, the
scapulo-thoracic muscles such as trapezius and serratus anterior
were found to be less active in the B–W–S compared to those of
the normal group.

3.3. GH contact force versus net moment regression analysis

Fig. 4A shows the GHF for the normal subjects against the net
moment obtained in this study and those obtained by Praagman
et al. (2000). The regression in this study was calculated (the

Pearson correlation=0.97) using the average of all Tasks is shown
in the following equation:

$$ F_{GH} = 9.41 + 37.8 [M_{GH}] $$

The same approach was taken for B–W–S; however, no strong
linear relationship (Pearson correlation =0.75) was found
between the B–W subjects’ net moment and their GHF (Fig. 4B).

4. Discussion

The inverse dynamic model of shoulder joint was used to
analyse and understand the biomechanical function of subjects
with the reversed anatomy B–W prosthesis during everyday
activities. The NSM has recently been used to analyse the
joint and muscle forces for 10 normal subjects (Charlton and
Johnson, 2006) while performing 10 ADL. Comparison of the six
tasks in common with that of Charlton and Johnson (2006)
shows a good agreement. They found that the largest force at
GH joint occurred while lifting a 2 kg weight to shoulder and
head height. In this study not all subjects were able to lift the 2 kg
to their head’s height so only 0.5 kg weights were used. The
highest GHF in this study was found for Task 8 (lifting 2 kg
shopping bag). The minimum and maximum forces observed
in this study and that of Charlton are compared and shown in
Table 3.

The results from this study highlight the importance of analysing
ADL tasks in comparison to standardised activities. In the earlier study
by Masjedi and Johnson (accepted for publication) it was found that
contact forces for abduction were greater for normal subjects than
for those with a B–W implant (Table 4). However, this was not the case
during the ADL tests. Furthermore comparison of the normal shoulder
data during standardised tasks (Charlton and Johnson, 2006) was
found to be in general agreement with those of Bergmann et al.
(2007) (who used an instrumented anatomical implant) as shown in
Table 4. However, during ADL (this study), when a larger external
force is acting on the arm, these results are substantially different.
Interestingly, in the ADL studies, results from B–W subjects were
similar to those of instrumented implant. The reported GHF for lifting
a coffee pot of 1.4 kg (Bergmann et al., 2007) reached a maximum of
103% of the body weight. For the B–W–S the maximum mean reached
the value of 110% of the body weight whereas for normal subjects this
value reached only the maximum mean value of around 62% of
the body weight. The moment arms of GH muscle in the B–W design
differ to those of normal subjects (Masjedi and Johnson, accepted for
publication) and hence similar to the anatomical prosthetic subjects.
The finding above therefore suggests that (if the modelling data are
assumed to be correct) GH muscles will be highly activated for the
prosthetic shoulders regardless of the implant type. Further
investigation using emg is required to evaluate the muscle
activations before and post surgery.

The most active muscle in the recorded activities across
normal subjects was supraspinatus which has an important role
to direct the GHF towards the glenoid cavity. In the absence of
supraspinatus, superior translation may occur that can cause
subacromial impingement. The B–W design is fully constrained
and therefore this translation can never take place. Deltoid was
found to be the most active muscle across activities for the B–W
group. Following the deltoid muscle, the coracobrachialis was
found to be the second most active muscle. The forces were also
found to be relatively large, with averages of 43 and 45 N (across
B–W–S and tasks), respectively, in comparison to 1.8 N for those
of normal subjects. Due to its moment arm, the coracobrachialis
muscle can act as an important flexor in the absence of cuff
muscles. The anterior parts of the deltoid have also relatively
large elevation moment arms in forward flexion and therefore are

Fig. 3. The regression in this study was calculated (the

$$ F_{GH} = 9.41 + 37.8 [M_{GH}] $$

(7)
mostly active in tasks involving high flexion moments such as Tasks 8 and 6. Since the deltoid is the main abductor for the B–W–S, their ability to perform the tasks probably is highly dependent on their muscle strength (PCSA). Finally, the decrease in activation of scapulo-thoracic muscles such as trapezius and serratus anterior for B–W–S is most likely due to a decrease in the range of humeral and therefore scapular motions in comparison to those of normal subjects. The validated biomechanical model can therefore be used in order to gain knowledge about the pathology as well as possible post surgical rehab for subjects with reversed shoulder replacement.

The relationship between the GHF and net moment is of great interest. If there were a consistent relationship, prediction of the GH contact forces would be straightforward. Using the NSM,
Charlton and Johnson (2006) followed the study by Praagman et al. (2000) who found a linear relationship between the GHF and the net moment predicted by the Delft Shoulder Model (DSM) (Van der Helm, 1991). The regression equations from each model were found to be in general agreement (Charlton and Johnson, 2006). The results from this study showed that DSM regression predicts slightly lower GHF than the regression made in this study using the NSM. This variation could be due to the difference in the activities tested, since, unlike this study, no additional external load was used in the study of Praagman et al. (2000). For the results predicted for the B–W–S, no linear relationship was found between the net moment and the contact force. In order to examine this further, the relationship between the GHF and net moment for simulated abduction task reported by Charlton and Johnson (2006) was analysed. It was found that for this task net moment and GHF have a linear relationship. Further investigation showed that the summation of the deltoid (anterior, posterior and middle) potential moment (PCSA-Momentarm) in this task has a linear relationship with the net moment. This identifies that the GHF is perfectly correlated with summation of the deltoid potential moment (the Pearson correlation was 0.98 and \( p < 0.05 \)). The reason for not observing a linear relationship for B–W–S is most likely to be that in the lack of RC muscles, when muscles such as middle deltoid with a favourable line of action reach their capacity, other muscles with less favourable moment arms and line of action (e.g. coracobrachialis) would become activated, and increase the GHF in an unpredictable manner.

Furthermore, the muscle forces differ substantially between the individual B–W–S due to a variation in the kinematics of these subjects (Masjedi and Johnson, 2008). For normal anatomy subjects however, the active muscles are almost perpendicular to the glenoid cavity and none of them reach their capacity.

The approach taken in this study here can be applied to different types of shoulder prostheses, for better understanding of the functionality of different designs. The predicted loading range can then be applied in the studies that are concentrating on the implant design and fixation (Chebli et al., 2008; Parsons et al., 2009) in order to optimise the design.

This study is only based on a single dataset and the muscles PCSA has not been scaled for the subjects. Imaging data are therefore required in future studies for a more accurate representation of the subjects. Future studies will also be able to take advantage of the newly developed instrumented implant (Taylor et al., 2008) which is capable of measuring in vivo loads. Once the data from the instrumented implant are available, the results from this study can be used for possible validation studies of the biomechanical models of the shoulder. These models can then be used for supporting of surgical procedures in order to achieve the optimal outcome for the patients.

Conflict of interest statement

Milad Masjedi and Garth R. Johnson, as the authors of the submitted paper have no financial and personal relationships with other people or organizations that could inappropriately influence (bias) this work.

Acknowledgments

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Table 3
Comparison between predicted GH contact force magnitude for normal subject in this study with those of Charlton and Johnson (2006).

<table>
<thead>
<tr>
<th>Study</th>
<th>Maximum GHF</th>
<th>Task</th>
<th>Minimum GHF</th>
<th>Task</th>
</tr>
</thead>
<tbody>
<tr>
<td>This paper</td>
<td>77% of body</td>
<td>Lifting shopping bag</td>
<td>29.8% body weight</td>
<td>Reach to opposite axilla</td>
</tr>
<tr>
<td></td>
<td>weight (572 N)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Charlton and Johnson</td>
<td>75% of body</td>
<td>Reaching to head height</td>
<td>33.57% body weight</td>
<td>Reach to opposite axilla</td>
</tr>
<tr>
<td>(2006)</td>
<td>weight (577 N)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 4
Comparison between predicted GH contact force magnitude for normal and B–W subject using the Newcastle shoulder model with those of Bergmann et al. (2007).

<table>
<thead>
<tr>
<th>B–W subjects NSM (%)</th>
<th>Normal subject NSM (%)</th>
<th>Bergmann (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Abduction 45°</td>
<td>34</td>
<td>45</td>
</tr>
<tr>
<td>Flexion 90°</td>
<td>62</td>
<td>83</td>
</tr>
</tbody>
</table>

Fig. 4. The GH contact force versus the net moment and the regression made for (A) normal subjects and (B) B–W subjects.
Appendix. Supporting information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2010.05.024.

References


