11th Conference of the International Shoulder Group
July 14 - July 16
Winterthur, Switzerland

Programme and Book of Abstracts

www.internationalshouldergroup.org
Welcome to Winterthur

Dear Conference Participants

The ZHAW Zurich University of Applied Sciences is very happy to host the 11th Conference of the International Shoulder Group ISG 2016.

The host city Winterthur is known as a very important place in Switzerland at which the industrial era was initiated in the 18th century. One of the first Technical Universities was founded 1874 in this place - known as the former “Technikum Winterthur” and now grown up to the Zurich University of Applied Sciences. The University represents an important basis for the numerous implant manufacturers founded in Winterthur.

We received around 40 scientific abstracts from a dozen countries. In addition, two interesting keynote lectures will complement the scientific sessions, and an openSim workshop will take part with approximately 30 participants. We are looking forward for an excellent and stimulating meeting in the light of shoulder research.

We would like to thank for the contribution of the numerous sponsors who support our conference. Please take your time and visit the industrial exhibition.

I hope that you enjoy your time in Winterthur and wish you a delightful conference.

Daniel Baumgartner

ISG 2016 Conference Organiser

Technikum at the time as Albert Einstein was teaching here (approx. 1901).
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Wi-Fi access:   Network:   ZHAW-event
                Password:   58ZHAW74event
Acknowledgements

Scientific Committee

We would like to thank the following members of the scientific committee for their assistance in reviewing the abstracts:

- Dimitra Blana
- Andrea Cutti
- Clark Dickerson
- Philippe Favre
- Bernd Heinlein
- Andreas Kontaxis
- Gretchen Oliver
- Didier Staudenmann
- Stefan van Drongelen
- Dirk Jan Veeger

Organising Committee

We would also like to thank the following people for their assistance with organising and running the conference:

- Daniel Baumgartner
- Lukas Gossweiler
- Roman Kuster
- Bruno Schmid
- Susanne Schütz

We additionally thank the International Society of Biomechanics for their ongoing support and the board of the International Shoulder Group for their contribution to organise the conference.
Sponsors

We would like to thank our Main Sponsor for his generous support.

We would also like to thank our Co-Sponsors for their contributions.

We would also like to thank the Award Sponsors:
Conference Program
Note that presentations in grey will probably not take place

Day 1: Thursday, July 14th 2016
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8400 Winterthur

13.30 - 18.00
OpenSim Workshop
Matias, Ricardo & Blana, Dimitra

18.00 - 19.30
Registration and Welcome Reception

Day 2: Friday, July 15th 2016
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8400 Winterthur

08.30
Registration and Coffee

09.00
Welcome
Baumgartner, Daniel

Session 1: Athletics and Sports
Chair: Gretchen Oliver & Ursina Arnet

09.15
Giovanardi, Andrea
Swimmer Hand Trajectory Using Wearable Inertial Magnetic Measurement Units: A Preliminary Study

09.30
Habechian, Fernanda
Children and Adolescents Competitive Swimmers, Amateurs and Non-Athletes: Scapular Kinematics and Relationship with Latissimus Dorsi Stiffness

09.45
Plummer, Hillary
The Influence of Lumbo-Pelvic Stability on Youth Pitching Kinematics

10.00
Pogetti, Livia Silveira
Relation between Core Endurance Time and Isokinetic Peak Torque of the Shoulder in Throwing Athletes with Shoulder Pain

10.15
Forgiarini Saccol, Michele
No Differences in Rotator Cuff Strength in Rugby Athletes with Previous Shoulder Injury

10.15
Keynote 1
PD Dr med Matthias A. Zumstein, University Hospital Berne (Switzerland)

11.00
Coffee Break & Visit of Industry Exhibition

Session 2: Arthroplasty
Chair: Andreas Kontaxis & Ilaria Parel

11.45
Favre, Philippe
Combined Experimental and Numerical Methods to Assess the Factors Affecting Primary Stability of a Stemless Humeral Prosthesis

12.00
Inyang, Adijat Omowumi
Investigating the Three-Dimensional Geometry of the South African Proximal Humerus

12.15
Schmid, Bruno
Comparison Between an Anatomic vs. Reverse Shoulder Prosthesis by Means of an Artificial Shoulder Simulator

12.30
Hopkins, Andrew
Longevity Gains for Screw Fixated Glenoid Baseplates through Navigation Guided Optimization

12.45
Kontaxis, Andreas
Pre-Operative Planning and Accurate Implantation Can Increase Impingement Free Range of Motion in Reverse Shoulder Arthroplasty – A Cadaveric Validation

13.00
Lunch
Session 3: Scapular Kinematics  
Chair: Dimitra Blana & Clark Dickerson

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<td>Parel, Ilaria</td>
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<td>Effects of Muscle Fatigue on Scapular Kinematics of Healthy Overhead Athletes during Simulated Throwing</td>
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15.15 Coffee Break & Visit of Industry Exhibition

Session 4: Methods / Modelling  
Chair: Paula Ludewig & Bernd Heinlein

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17.00-18.30 Open Lab Session (guided Tour with 5 events, 15 Min each)

19.00 Conference Dinner at Restaurant Gate 27, Theaterstrasse 27b, 8400 Winterthur

Day 3: Saturday, July 16th 2016  
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8400 Winterthur

Session 5: Functional Assessment and Pain  
Chair: Stefan van Drongelen & Roman Kuster

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<td>09.30</td>
<td>Lang, Angelica</td>
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<td>Mancuso, Matteo</td>
<td>A novel Virtual Reality Based Approach towards the Instrumented Dunctional Test of the Shoulder</td>
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<td>10.00</td>
<td>Barreto, Rodrigo</td>
<td>Bilateral MRI Findings in Individuals with Unilateral Shoulder Pain</td>
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10.15 Coffee Break & Visit of Industry Exhibition
Session 6: Rotator Cuff  
Chair: Philippe Favre & Andrea Giovanni Cutti

10.45  Kuster, Roman Peter  
Experimental Shoulder Testing under in-vivo Anatomy and Physiology

11.00  Kolk, Arjen  
Glenohumeral Elevation is Reduced in Massive Posterosuperior Rotator Cuff Tears: a 3D Motion Analysis in Rotator Cuff Disease

11.15  Lawrence, Rebekah  
The Effect of Humeral Elevation and Retroversion on Supraspinatus Subacromial Compression During a Simulated Reaching Task

11.30  Ludewig, Paula  
Mechanical Internal Impingement of the Supraspinatus Tendon During a Simulated Reaching Task

12.00  Lunch

Session 7: Muscle  
Chair: Dirkjan Veeger & Adijat Omowumi Inyang

13.00  Cudlip, Alan  
Examining NSCA and ‘Powerlifting’ Bench Press Techniques on Upper Extremity Muscle Activity

13.15  Kim, Soo  
Electromyographic Investigation of Anterior and Posterior Regions of Supraspinatus: A Novel Protocol Based on Anatomical Insights

13.30  Oliver, Gretchen  
Implication of Gluteus Medius Fatigue on Overhead Throwing Kinematics

13.45  Vidt, Meghan  
Spatial and Load Dependency of Upper Limb and Shoulder Muscle Activity During Submaximal Exertions

14.00  Wochatz, Monique  
Scapular Muscle Activity Pattern during Isokinetic Shoulder Flexion and Extension in Passive and Maximum Effort Conditions

14.15  ISG General Assembly

Session 8: Poster Presentations: (2 Min)

15.15  Heinke, Lars  
Evaluation of Passively Induced Shoulder Stretch Reflex Using an Isokinetic Dynamometer in Men

15.17  Baumgartner, Daniel  
Experimental Analysis of the Subscapularis Force Response due to an External Load Impact

15.19  Sarshari, Ehsan  
A Framework for Forward-Dynamics Simulation of the Human Shoulder

15.21  Springer, Jochen  
Simulation of Forces in the Glenohumeral Joint with a Multi Body Simulation Model of the Human Shoulder

15.23  Bossuyt, Fransiska  
Shoulder Pain in Individuals with Spinal Cord Injury

15.25  Cutti, Andrea  
Proposal of a Method to Summarize Quantitative Motion Analysis Data of the Upper Limb for Clinical Applications

15.27  Kuster, Roman Peter  
Biomechanical Analysis of Roller Skiing to determine the Force Profile applied on Ski Poles

15.29  Staker, Justin  
Three-Dimensional Kinematics of Shoulder Laxity Examination: Reliability, Validity and Relationship

15.31  Scheunert, Erika  
Assistive System of the Shoulder Joint From Complex Exoskeleton to Functional 1 DoF Support

15.33  Smith, Simon  
Total Shoulder Replacements Wear Tested in a Unique Shoulder Simulator

15.45  Poster Session and Industry Exhibition with Wine & Cheese

16.45  Award Ceremony and Closing Remarks

Day 4: Sunday, July 17th 2016

07.30  Expedition to the Swiss Alps, one-day trip (including a 2 hrs walk)  
Keynote speakers

PD Dr Matthias A. Zumstein

Matthias Zumstein is the Head of Shoulder, Elbow and Orthopaedic Sports Medicine at the University Hospital of Berne and Associate Professor at the University of Berne. He was a professional Handball player and represented Switzerland at the Atlanta Olympics in 1996. His research interests are predominantly in shoulder and elbow with a particular interest in the basic science of the rotator cuff muscles. He was awarded with the Charles S. Neer award in the basic science category in 2007 with the work "Reduced Muscle Degeneration and Decreased Fatty Infiltration after Rotator Cuff Tear in a PARP-1 Knock-Out Mouse Model". He is currently a board member of Swiss Orthopaedics and the European Shoulder and Elbow Society as well as associate member of the ASES.

Current Clinical Aspects in Shoulder Surgery (Friday, 10.15)
Clinical aspects in shoulder surgery are the basis to determine essential scientific research programmes. It is therefore of high interest for the biomechanical research to know about relevant clinical questions. The keynote deals with clinical and surgical challenges, which could be proven by interactive collaborations between Hospitals and Universities.

Prof Dr Tobias Nef

Tobias Nef is the Head of the Gerontechnology & Rehabilitation ARTORG Centre for Biomedical Engineering Research in Berne. He is a specialist in robot assisted upper extremity rehabilitation for stroke patients. Within his thesis "ARMin - Multimodal Robot for the Movement Therapy of the Upper Extremities" he developed a rehabilitation device that was honored with several awards such as the Swiss Technology Award 2006 and the Hans-Eggenberger Award 2008.

The „Shoulder Challenge“ in Upper Limb Rehabilitation Robotics (Saturday, 08.30)
Robot-assisted arm training has become popular for the rehabilitation of brain injured patients. A large number of arm robots have been developed and some of them are currently used in daily clinical practice. In this presentation, an overview of the current state of the art will be given with a special focus on different shoulder actuation principles.
Abstracts

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Swimmer hand trajectory using wearable inertial magnetic measurement units: a preliminary study

1Andrea Giovanardi, 2Silvia Fantozzi, 3Matteo Cortesi, 4Maria Luisa Ruspi, 5Giorgio Gatta,
1School of Pharmacy, Biotechnology and Sport Science, University of Bologna, Bologna, Italy
2Department of Electrical, Electronic, and Information Engineering, University of Bologna, Bologna, Italy
3Department for Life Quality Studies, University of Bologna, Bologna, Italy

INTRODUCTION
The arm-stroke technique has a key role in the swimming performance. Although the forward motion of the swimmer is due to the lower and upper limbs, the latter ones produce the most part of propulsion. The features of the underwater hand trajectory seem to be highly correlated with the swimmer velocity [1]. Previous studies investigated the hand trajectory, traditionally using underwater cameras [2, 3]. However, this type of analysis involves several drawbacks, like i) time-consuming installation, calibration and data processing, ii) a limited field of view that allows the analysis of few strokes, and iii) turbulences, refraction of light and parallax error at the water-air interface [3]. To overcome these limits of video analysis, in the last ten years, wearable inertial magnetic measurements units (IMMUs) were exploited for the kinematic analysis of swimming [4, 7]. These devices, being directly fixed on the swimmer, allow a continuous data acquisition during the whole swim. Swimming spatio-temporal descriptors, joint kinematics, and body roll angles were widely investigated in literature [3, 4, 7]. On the contrary, though the hand trajectory is a key indicator of good technique and performance, only one study addressed this issue using IMMUs but without providing extensive validation [5]. Moreover, to date, the evaluation of the hand trajectory is left to the qualitative assessment of coaches during workout. Thus, the aim of this study was to quantitatively estimate the hands trajectories of swimmers using IMMUs, and to validate this estimation in dry land condition using a stereophotogrammetric system as gold standard.

METHODS
Protocol
From a biomechanical point of view, each side was modeled as an open kinematic chain constituted by thorax, upper-arm and forearm with 5 degrees of freedom. Similarly to the representation described by Cutti et al. [6], the shoulder was considered as the ball-and-socket joint between thorax and arm, while the elbow was considered as the double-hinge joint (with non-intersecting axes) between arm and forearm. For each segment that formed 2 joints, both a proximal and a distal embedded anatomical reference systems (ARS) were defined. ARSs definitions introduced by Cutti et al. [6] was adopted apart from the facts that: (1) the static calibration acquired for the definition of the thorax ARS was performed with the subject lying still; and (2) the proximal forearm ARS was rotated of -90° along the Y-axis because the elbow joint during swimming is almost completely pronated in many phases of a stroke [7]. The positioning of IMMUs on body segments is shown in Fig. 1 (left).

Figure 1. Left: IMMUs positioning on body segments detailing the anatomical systems of reference axes. Right: lateral and anterior views during front crawl simulation.

The protocol required three types of calibration tasks, as described in [6, 7], that allowed to compute each ARS with respect to the corresponding IMMU technical system of reference, for each dynamic task. The protocol provided accurate estimation of 3D joint angles in simulated swimming [7]. The hand trajectory was estimated considering the kinematic chain of the rigid body segments up to the midpoint between the styloids, using as distances the coordinates of II, GH, the midpoint between the humerus epicondyles and the midpoint between the styloids, measured in the static calibration trial.

Test Validation
One male swimmers (Age: 34 years; Height 180 cm; Mass 80 Kg; Training 18 years) agreed to participate and freely signed the informed consent. The swimmer was asked to swim in the same way he would have done in a swimming pool (Fig.1 Right). Thirty complete front crawl arm-stroke cycles were available after data collection. Data collection was performed using the APDM Opal system as the

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IMMUs system (5 nodes, 128Hz) together with a stereophotogrammetric system (BTS SMART-DX 7000, 7 cameras, 250Hz) used as the gold standard. To compare kinematic data estimated from both systems, 5 clusters (four 10-mm markers and one IMMU attached onto a rigid light-weighted wooden plate) were built and firmly fixed onto the swimmer body. Anatomical C.A.S.T. calibration for humerus epicondyles and forearm styloids of both limbs were performed. Additional markers were placed also in the thorax anatomical landmark (IJ, PX, C7, and T8).

Data Analysis
Three different hand trajectories were computed: 1) the trajectories directly measured with the stereophotogrammetric system by means of markers (SPSM), considered as the gold standard; 2) the trajectories estimated from stereophotogrammetric data applying the kinematic chain model (SPSE); and 3) the trajectory estimated from IMMUs data applying the same model (IMMUE). The hand trajectories were computed in the thorax reference system, to have a common reference. Thorax X axis was pointing caudally, Y axis pointing to the left of the subject, Z axis pointing frontally (downward during swimming).

Comparison among the 3 methods was performed analyzing the root mean square error (RMSE), Pearson product-moment correlation coefficient (r), Bland Altman mean and coefficient of repeatability of the x, y, z components of the hands trajectories.

RESULTS
Overall results of the comparison of the 3 methods are shown in Table 1, while an example of the mean right hand trajectory is illustrated in Figure1.

<table>
<thead>
<tr>
<th></th>
<th>RMSE (cm)</th>
<th>r</th>
<th>BA Mean (cm)</th>
<th>BA CR (cm)</th>
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<td>SPSM vs SPSE</td>
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<tr>
<td>Median</td>
<td>8.4</td>
<td>0.99</td>
<td>-7.9</td>
<td>6.8</td>
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<tr>
<td>IQR</td>
<td>2.5</td>
<td>0.04</td>
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<tr>
<td>SPSM vs IMMUE</td>
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<tr>
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<td>0.94</td>
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<tr>
<td>IQR</td>
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<td>9.0</td>
<td>2.6</td>
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<tr>
<td>SPSE vs IMMUE</td>
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<tr>
<td>Median</td>
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<td>2.2</td>
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Table 1. Median and Inter quartile range (IQR) values for RMSE, r, Bland-Altman Mean and Bland-Altman coefficient of repeatability (BA CR) for each comparison.

DISCUSSION AND CONCLUSIONS
Comparing the two estimated hand trajectories with those measured by the gold standard, similar and shifted patterns were observed (no shift was observed when comparing SPSE vs IMMUE). This error can be explained by the rigid body model assumption that could be critical for the thorax segment. Future developments will take into account also the motion of the gleno-humeral joints, using 2 additional IMMUs on the scapula, now intentionally omitted to avoid problems related with their fixation, which could increase the drag forces for the swimmer. However, the high correlation values found between methods enhanced the similarity among the 3 hand trajectories patterns. The accuracy reached with the estimation can be considered sufficient to satisfy the coaches and athletes training purposes.

REFERENCES
INTRODUCTION

The swimming practice in competitive level is increasing in children and adolescents, as well as the number of shoulder injuries in this population [1,2]. Madsen et al. (2011) [3] have found scapular dyskinesis in pain-free adolescents and adults swimmers. It is important to understand scapular kinematics in swimmers because appropriate scapular movement is essential for them to perform their strokes with the adequate range of motion. In addition, literature has already shown that one of the factors that can alter scapular kinematics pattern in adult swimmers is the latissimus dorsi stiffness [4]. This muscle is one of the most activated during the swimming, assisting the body propulsion in the water [5]. It has attachments that pass on the inferior angle of the scapula, which could in the event of the muscle being stiff, due to its overuse, limit scapular kinematics [4]. It would be interesting to know if these alterations are also present in children and adolescents swimmers. This could help in understanding their adaptations and biomechanical aspects due to the swimming practice. The comprehension of these aspects could also help in developing adequate evaluation, treatment and prevention of shoulder injuries, which could get worse in the future.

The primary aim of this study was to analyze scapular kinematics in children and adolescents competitive swimmers, amateurs and non-athletes. The secondary aim was to check if there is a relationship between the latissimus dorsi stiffness and scapular kinematics in competitive swimmers.

METHODS

Thirty-nine children/adolescents were evaluated and divided into 3 groups: 13 competitive swimmers (11.84±2.11 years; 48.62±11.21 kg; 1.54±0.10 m), 13 amateurs (11.69±1.79 years; 43.53±10.22 kg; 1.52±0.12 m) and 13 non-athletes (10.30±1.65 years; 41.65±13.38 kg; 1.46±0.11 m). All participants had no history of shoulder/cervical dysfunction and at least 150° of arm elevation. The competitive swimmers had to practice for a minimum of 3 years, at least 3 times per week (minimum of 4,000 m/day) and they had to participate in professional level competitions for at least 1 year. The swimmers amateurs had to practice for a minimum of 1 year, maximum 2 times per week. The non-swimmers could not be involved in the practice of any sports involving the upper limb.

Scapular kinematics was analyzed during elevation of the arm in the scapular plane and assessed with the electromagnetic tracking device Flock of Birds® integrated with the MotionMonitor™ software. Local coordinate systems were established using the digitized landmarks following the International Society of Biomechanics recommended protocol [6]. All individuals had the dominant arm evaluated. Three repetitions of arm elevation were performed. Scapular kinematics was analyzed at 30°, 60°, 90° and 120°. The data were averaged over the 3 repetitions.

The latissimus dorsi stiffness was tested through a digital inclinometer (OE-220, ITO – Physiotherapy & Rehabilitation, Japan). The inclinometer was placed 5cm below the inferior angle of the scapula, in the dominant side, with the subject in prone position and hand below the front. Three trials were performed. Shapiro-Wilk test was used to check the data normality. A 2-way mixed model analysis of variance (ANOVA) was used for the 3 scapular rotations, in separate, considering humeral angle (30°, 60°, 90° and 120°) as within factor and group (competitive swimmers, amateurs and non-swimmers) as between factor. The Bonferroni test for post hoc analysis was used when necessary. A Pearson correlation test was performed to check the relationship between the scapular kinematics and the latissimus dorsi stiffness. A p <0.05 was considered significant.

RESULTS

A significant angle X group interaction was found for the scapular internal rotation at 120° (p=0.009), with competitive swimmers showing more internal
rotation compared to non-athletes (mean difference: 9.3°) (Figure 1). For scapular upward rotation and scapular tilt no differences were found among groups ($p>0.05$).

Regarding the relationship, no significant correlation was found between the latissimus dorsi stiffness and the scapular kinematics in the competitive swimmers for any of the scapular kinematics in any of the four angles of arm elevation ($p>0.05$).

![Figure 1. Means of scapular internal/external rotation. The error bars represent the standard error of the mean. *Significant difference ($p<0.05$) between competitive swimmers and non-athletes.](image)

DISCUSSION AND CONCLUSIONS
According to literature, the scapula normally tends to externally rotate at the end of arm elevation in healthy subjects [7]. However, in the present study, the competitive swimmers presented higher scapular internal rotation compared to the non-athletes at 120°.

The humeral internal rotation is a movement required during the swimming practice and, consequently, the scapular internal rotation [8]. This may explain the results found. However, even being an important movement during the swimming performance, literature has already shown that increased scapular internal rotation is associated to injuries [9]. Considering this fact, the competitive swimmers that participated in the present study may be predisposed to develop shoulder injury in the future. Regarding the latissimus dorsi stiffness, no relationship was found with the scapular kinematics in competitive swimmers. This result diverges from Laudner and Williams (2013) [4] that found moderate to good relationship between increased latissimus dorsi stiffness with increased scapular upward rotation and posterior tilt and decreased internal rotation. This could be explained by the fact that stiffness may be an adaptation that occurs along time.

It is possible to conclude that the swimming practice (in a competitive level) seems to affect the scapular internal rotation in the competitive swimmers of the present study. However, no relationship was found between scapular kinematics in competitive young swimmers and latissimus dorsi stiffness. A more comprehensive understanding of scapular kinematics in the young population of competitive swimmers can help in the development of appropriate evaluation, treatment and preventive protocols for this population.

REFERENCES

ACKNOWLEDGEMENTS
This study was supported by Fundação de Amparo à Pesquisa do Estado de São Paulo (2013/19711-4).
INTRODUCTION

Baseball pitching is a dynamic upper extremity skill that requires coordination of the upper and lower body to produce optimal mechanics for the most efficient ball release. During pitching, energy is generated from the lower extremities, pelvis and trunk and transferred to the shoulder and then the hand. Therefore, stability at the pelvis and trunk is important for the efficient generation and transfer of energy to the upper extremity. A lack of trunk and pelvic stability could result in increased loads at the upper extremity.

Clinical tests aimed at assessing athletes’ movement patterns can be used to identify the potential for injury risk. One common screening test that can be used to assess lumbo-pelvic stability is the single leg squat (SLS). This test can be used to detect instability that may predispose one to altered upper extremity mechanics.

The purpose of this study was to classify lumbo-pelvic stability during a SLS and determine how these results relate to pitching mechanics. It was hypothesized that pitchers with poor stability scores would exhibit altered pelvis and trunk lateral flexion, trunk flexion, shoulder elevation, and shoulder plane of elevation during pitching compared to those with higher stability scores.

METHODS

Participants

Sixty-four youth baseball pitchers (11.19 ± 1.50 years; 154.23 ± 10.75 cm; 48.72 ± 11.08 kg) participated. Informed consent and parental assent was obtained prior to testing.

Procedures

The MotionMonitor™ (Innovative Sports Training, Chicago, IL) synched with an electromagnetic tracking system (Track Star, Ascension Technologies Inc., Burlington, VT) was used to collect data. Ten electromagnetic sensors were placed at the following locations: (1) trunk at C7; (2) pelvis at S1; (3) throwing arm humerus; (5) throwing arm forearm; (6-7) bilateral shank; (8-9) bilateral femur; (10) stride leg third metatarsal. All kinematic data were sampled at a frequency of 100 Hz. Raw data were independently filtered along each global axis using a 4th order Butterworth filter with a cutoff frequency of 13.4 Hz. Raw data regarding sensor orientation and position were transformed to locally-based coordinate systems for each respective body segment. Euler angle decomposition sequences were used to describe both the position and orientation. International Society of Biomechanics standards and conventions were used to define pelvis, trunk and shoulder kinematics.

Participants performed a SLS on their non-throwing, stride leg. Participants placed their hands on their hips and flexed their contralateral knee. They were instructed to squat as low as possible before returning to a fully extended knee. The descent phase (initiation of knee flexion to maximum knee flexion) was selected for analysis. A 3-point rating scale was used to categorize trunk lateral flexion and pelvis lateral flexion during the SLS as having significant (0), mild (1), or no movement (2). Higher scores indicated an optimal movement quality. For both trunk and pelvis measures, 0 = movement >10° in any direction; 1 = movement that was 6-10° in any direction; 2 = movement < 5° in any direction.

Following the SLS, data were collected for five accurate four-seam fastball pitches. An accurate pitch was defined as a pitch passing through the strike zone and was determined by a trained investigator. Participants threw from their age-regulated pitching distance (46ft/14.02m).

The fastest pitch was selected for analysis. Two, one-way analysis of variances (ANOVA) were performed to examine if pelvis and trunk lateral flexion, trunk flexion, shoulder elevation, and shoulder plane of elevation, at maximum shoulder external rotation (MER) of the pitch, were significantly different between trunk and pelvis stability score groups.
**RESULTS**

The mean pitch speed was 48.72 ± 7.70 mph (21.78 ± 3.44 m/s). The results are presented in Table 1. No significant differences in pitching kinematics were observed when grouped by pelvis stability score. Trunk lateral flexion, towards the non-throwing arm, was significantly greater in pitchers with a trunk stability score of 0 compared to 2 (p = 0.03).

**DISCUSSION AND CONCLUSIONS**

The results of this study indicate that pitchers who exhibited poor lumbo-pelvic stability during a SLS had altered pitching mechanics. Specifically, greater trunk lateral flexion was observed in pitchers with poor trunk stability scores, which may be indicative of a decreased ability to control trunk position during pitching. Trunk lateral flexion helps to position the shoulder in the proper arm slot in preparation for release. Increased lateral flexion may alter energy transfer to the shoulder and result in increased kinetics. The SLS may be a useful tool for clinicians to assess trunk stability in pitchers as those with poor stability may also have altered trunk lateral flexion during pitching.

There are several limitations of this study. The sample size of pitchers was small and was then further divided into three groups based on stability score. This led to an analysis of unbalanced groups, specifically for trunk stability where few participants achieved the maximum score. Additionally, knee flexion during the SLS was not standardized between participants. This could have affected the amount of perturbation and thus the degree of lumbo-pelvic stability required during the SLS.

Future research should aim to characterize the risk of poor trunk and pelvis stability on upper extremity injury in baseball players. As well as defining muscular strength impairments that characterize poor stability, which will provide critical knowledge that can be used to develop training and preventative rehabilitation program aimed at reducing injury risk.

**REFERENCES**

1. Bunn JW. *Scientific Principles of Coaching*, 1972

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Table 1. Comparison of pitching kinematics, at maximum shoulder external rotation, between pitchers classified by pelvis and trunk stability scores. Mean (SD).

<table>
<thead>
<tr>
<th>Stability Score</th>
<th>Pelvis Lateral Flexion</th>
<th>Trunk Lateral Flexion</th>
<th>Trunk Flexion</th>
<th>Shoulder Elevation</th>
<th>Shoulder Plane of Elevation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0 (n=21)</td>
<td>-2.52 (5.77)</td>
<td>-20.08 (11.04)</td>
<td>-1.52 (11.16)</td>
<td>-89.22 (15.31)</td>
<td>8.54 (21.95)</td>
</tr>
<tr>
<td>1 (n=21)</td>
<td>-3.35 (6.44)</td>
<td>-18.29 (16.55)</td>
<td>-2.32 (12.11)</td>
<td>-81.64 (12.38)</td>
<td>11.31 (21.56)</td>
</tr>
<tr>
<td>2 (n=22)</td>
<td>-1.32 (7.56)</td>
<td>-15.31 (13.22)</td>
<td>2.59 (6.97)</td>
<td>-87.01 (10.23)</td>
<td>11.76 (19.96)</td>
</tr>
<tr>
<td>Trunk</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0 (n=35)</td>
<td>-3.41 (6.74)</td>
<td>-22.11 (14.01)*</td>
<td>-2.45 (10.88)</td>
<td>-86.61 (15.32)</td>
<td>9.85 (17.76)</td>
</tr>
<tr>
<td>1 (n=20)</td>
<td>-0.07 (11.06)</td>
<td>-14.33 (11.06)</td>
<td>1.05 (9.96)</td>
<td>-83.63 (9.32)</td>
<td>14.55 (20.34)</td>
</tr>
<tr>
<td>2 (n=9)</td>
<td>-3.13 (7.88)</td>
<td>-9.15 (12.63)*</td>
<td>4.55 (7.23)</td>
<td>-88.70 (9.79)</td>
<td>4.46 (31.97)</td>
</tr>
</tbody>
</table>

Pelvis and trunk lateral flexion (−) is towards non-throwing arm; Trunk flexion (−) is extension; Shoulder elevation (−) is elevation; Shoulder plane of elevation (+) is anterior to frontal plane. *Indicates significant difference (p<0.05).
Relation between Core Endurance Time and Isokinetic Peak Torque of the Shoulder in Throwing Athletes with Shoulder Pain

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1Laboratory of Analysis and Intervention of the Shoulder Complex, Federal University of São Carlos, São Carlos, Brazil
2Physical Therapy, UNINORTE – University Center of North/Laureate International Universities (LIU), Manaus, Brazil

INTRODUCTION
Integrated action of the kinetic chain is essential to provide proper function of the shoulder in throwing athletes. As the shoulder is the receiver and transmitter of forces coming from the core[1,2], alterations in any of the components of the chain may contribute to the occurrence of dysfunction in the shoulder. According to Kibler[3], a decrease of 20% of the kinetic energy delivered from the core to the arm requires an increase of 80% in mass or 34% in rotational velocity at the shoulder to deliver the same amount of resultant force to the hand. This situation possibly predisposes the shoulder to injury. Therefore, the aim of this study was to investigate if there was correlation between the core endurance time and the isokinetic peak torque of the shoulder in throwing athletes with shoulder pain.

METHODS
Twenty-one throwing athletes with shoulder pain (6 females and 15 males, 21.57±2.89 years, 73.38±11.54 kg, 173.95±9.56 cm) participated in the study. They were handball (n=17), baseball (n=2) and softball (n=2) athletes at the college level. For the core endurance time, the trunk flexors, extensors, and lateral flexors were evaluated on the ipsilateral and contralateral sides of the dominant arm, and recorded in seconds. The trunk extensor and lateral flexor endurance tests required the athletes to be positioned with the pelvis over the edge of the treatment table and legs stabilized by two straps. At the beginning of the tests, the upper limbs were crossed over the chest and the upper body was lifted until the upper trunk was positioned horizontal to the floor. The athletes were instructed to keep the horizontal position as long as possible. The trunk flexor endurance test required the athletes to sit on the treatment table and place the upper body against a wedge with an angle of 60°. Both the knees and hips were flexed 90°. The arms were folded across the chest with the hands placed on the opposite shoulder and toes were placed under straps. Individuals were instructed to maintain the body position while the supporting wedge was pulled back 10cm to begin the test. The test ended when the upper body fell below 60° angle. Isokinetic concentric peak torque of the dominant arm (throwing arm) was also analyzed by the average of 5 maximal repetitions of internal and external rotation of the shoulder at 90°/s, 180°/s and 240°/s. For this test, the participants were evaluated in seated position while secured with pelvic and diagonal straps for trunk stabilization. The dominant arm positioned at 90° of shoulder abduction and 90° of elbow flexion. The olecranon was aligned with the dynamometer’s mechanical axis of rotation.

RESULTS
Table 1 brings all the results. Moderate positive correlation was observed between endurance time of the trunk flexors and shoulder external rotation peak torque at 90°/s, 180°/s and 240°/s. Weak positive correlation was found between endurance time of the trunk flexors and shoulder internal rotation peak torque at 90°/s and 180°/s, and moderate positive correlation at 240°/s. Weak positive correlation was demonstrated between endurance time of the ipsilateral trunk flexors and shoulder internal rotation peak torque at 180°/s. No significant correlations were found for the others variables.

DISCUSSION AND CONCLUSION
The results showed positive correlation of trunk flexors and ipsilateral trunk flexors endurance time with shoulder rotation isokinetic peak torque in throwing athletes with shoulder pain. These results are in agreement with Endo and Sakamoto[4] who also observed decreased core stability during bridge tests in baseball players with shoulder pain. The flexors and lateral flexors muscles serve as a vital component of the core[1]. These muscles increase the intra-abdominal
pressure making a hoop together with the thoracolumbar fascia, promoting functional stability of the lumbar spine and maximizing the force to throwing.[5] Poor core stability seems to increase the upper extremity risk of injuries in throwing athletes due to the overload that may be generated in the shoulder muscles. This fact may negatively affect the athletic performance. In conclusion, throwing athletes with shoulder pain showed positive correlation between endurance time of the trunk flexors and ipsilateral trunk flexors with shoulder rotation isokinetic peak torque. This suggests that the core stability should be considered in the rehabilitation program.

Table 1. Correlation between core endurance time and shoulder rotation isokinetic peak torque.

<table>
<thead>
<tr>
<th></th>
<th>r</th>
<th>p</th>
</tr>
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<tbody>
<tr>
<td><strong>Trunk flexors endurance</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ER 90°/s</td>
<td>0.75</td>
<td>0.001*</td>
</tr>
<tr>
<td>IR 90°/s</td>
<td>0.45</td>
<td>0.04*</td>
</tr>
<tr>
<td>ER 180°/s</td>
<td>0.68</td>
<td>0.001*</td>
</tr>
<tr>
<td>IR 180°/s</td>
<td>0.48</td>
<td>0.03*</td>
</tr>
<tr>
<td>ER 240°/s</td>
<td>0.59</td>
<td>0.001*</td>
</tr>
<tr>
<td>IR 240°/s</td>
<td>0.48</td>
<td>0.001*</td>
</tr>
<tr>
<td><strong>Trunk extensors endurance</strong></td>
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<td></td>
</tr>
<tr>
<td>ER 90°/s</td>
<td>0.08</td>
<td>0.72</td>
</tr>
<tr>
<td>IR 90°/s</td>
<td>0.31</td>
<td>0.18</td>
</tr>
<tr>
<td>ER 180°/s</td>
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<td>0.38</td>
</tr>
<tr>
<td>IR 180°/s</td>
<td>0.24</td>
<td>0.29</td>
</tr>
<tr>
<td>ER 240°/s</td>
<td>0.17</td>
<td>0.47</td>
</tr>
<tr>
<td>IR 240°/s</td>
<td>0.23</td>
<td>0.34</td>
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<tr>
<td><strong>Ipsilateral trunk flexors endurance</strong></td>
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<td></td>
</tr>
<tr>
<td>ER 90°/s</td>
<td>0.23</td>
<td>0.32</td>
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<tr>
<td>IR 90°/s</td>
<td>0.36</td>
<td>0.18</td>
</tr>
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<td>ER180°/s</td>
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</tr>
<tr>
<td>IR180°/s</td>
<td>0.46</td>
<td>0.04*</td>
</tr>
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<td>ER 240°/s</td>
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</tr>
<tr>
<td>IR 240°/s</td>
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<td>0.45</td>
</tr>
<tr>
<td><strong>Contralateral trunk flexors endurance</strong></td>
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<td></td>
</tr>
<tr>
<td>ER 90°/s</td>
<td>0.08</td>
<td>0.72</td>
</tr>
<tr>
<td>IR 90°/s</td>
<td>0.05</td>
<td>0.82</td>
</tr>
<tr>
<td>ER 180°/s</td>
<td>0.10</td>
<td>0.68</td>
</tr>
<tr>
<td>IR 180°/s</td>
<td>0.24</td>
<td>0.30</td>
</tr>
<tr>
<td>ER 240°/s</td>
<td>0.10</td>
<td>0.68</td>
</tr>
<tr>
<td>IR 240°/s</td>
<td>-0.06</td>
<td>0.77</td>
</tr>
</tbody>
</table>

ER, external rotation; IR, internal rotation; r, correlation coefficient value; *p<0.05.

REFERENCES

ACKNOWLEDGEMENTS
This work was supported by FAPESP (2014/09806-0).
INTRODUCTION

Rugby is a popular contact sport where the gameplay involves interrupting the progress of the opponent by knocking him down to the ground. This is known as tackle and it is the phase of game that half of the injury cases occur [1].

A ‘shoulder tackle’ is a technique in which the tackler engages the ball carrier with the shoulder [2] predisposing upper extremities structures to contact trauma and also to falling onto the ground generating excessive forces [3].

The epidemiology of rugby lesions demonstrated that upper extremities injuries has been estimated at 9.84 injuries per 1000 athletic exposures [4] and shoulder represents up to half of these upper limb lesions [3]. These shoulder lesions may involve a number of structures about the joint with acromioclavicular and glenohumeral joints the most frequently injured [1].

Considering the frequency of upper limb injuries in rugby players, it is expected that a number of athletes continues to play even with a previous shoulder lesion. Although these injuries may not represent a career-ending for athletes, they could compromise the shoulder performance in sport and also the rotator cuff strength.

Athletes with shoulder injuries had weakness of shoulder rotators [5-7] and even no shoulder strength alterations [7]. However, to the best of our knowledge, there are no studies that evaluated shoulder strength in rugby players with previous shoulder injuries. The aim of this study was to evaluate the shoulder rotator strength in rugby athletes comparing players with and without previous shoulder injury.

METHODS

Thirty five male rugby players participated of the study. Athletes answered a questionnaire about shoulder dominance, competitive level and previous injuries and surgeries. To participate in the study, athletes should practice rugby at least for two years, has no pain or upper limb symptoms that limited the practice of the sport in the previous three months and no previous upper limb surgeries. Based on the eligibility criteria, four athletes were excluded from the study.

Sixteen athletes without previous shoulder injuries (control group, CG) and fifteen athletes with previous shoulder injuries (injured group, IG) participated of the study. Table 1 presents participants characteristics.

Table 1. Descriptive characteristics of control and previous shoulder injured group of rugby players. Data are presented as mean (standard deviation).

<table>
<thead>
<tr>
<th></th>
<th>Control group (n=16)</th>
<th>Injured group (n=15)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>22.43 (4.35)</td>
<td>23.53 (5.66)</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>92.2 (26.06)</td>
<td>89.09 (10.09)</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>176.33 (8.86)</td>
<td>179.28 (5.34)</td>
</tr>
<tr>
<td>Rugby experience</td>
<td>7.93 (3.76)</td>
<td>8.66 (3.08)</td>
</tr>
</tbody>
</table>

For shoulder strength evaluation, a handheld dynamometer (Microfet 2 HHD, Hoogan Health Industries, West Jordan, UT, USA) was used. The athlete was sited in a chair and the evaluated shoulder was maintained with 30° abduction in the scapular plane, neutral rotation, 90° elbow flexion and forearm in the neutral position. The dynamometer was fixed to a wall support in order to avoid any interference in stabilization (Figure 1).

Figure 1. Shoulder rotation isometric strength test position.

The athlete performs shoulder rotator isometric strength against a handheld dynamometer positioned 3 cm proximal to the styloid process of the radius. All participants practiced twice to become familiarized with the procedure and the test was performed three
times for internal rotation and external rotation with 5 seconds of contraction and 30 seconds of rest between each measurement. The mean of the three strength tests was used for analysis purposes. The variables of interest were peak isometric strength of the internal and external rotators normalized to body weight (f).

Student’s t-test was used to compare the injured and dominant limb and also dominant and non-dominant shoulders (significance level of 5%).

RESULTS
Groups were not different regarding age, mass, height and sports experience.
No significant differences in internal or external rotation strength were found between dominant shoulder of control group and the previous injured shoulder in injured group (p=0.760 and 0.435, respectively).
Control group presented difference in external rotation strength comparing dominant and non-dominant shoulder (p=0.039).

DISCUSSION AND CONCLUSIONS
Rugby athletes that continues to play even with previous shoulder injuries did not present shoulder rotators strength differences compared to healthy rugby controls. We believe that the maintenance of a regular training regimen in those athletes with shoulder injuries may have contributed to this finding.
Also, shoulder injuries cause different impairments in sports performance and this can be related to shoulder overhead demands on each sport. Besides tackle contact, rugby athletes also needs shoulder movement to perform passes, so they can throw the ball lateral and backward [8]. However, considering the demand of shoulder stability on this movement, it can be expected that even with a previous shoulder injury, an athlete can play with a certain level of skill.
Considering the non-injured group, only external rotation strength on dominant shoulder was higher than in non-dominant shoulder. A previous study with shoulder rotator strength in rugby athletes with no history of shoulder pathology did not find strength imbalances in external or internal rotators [9]. Those authors also raised the question about the relevance of muscle imbalances in sports that traumatic shoulder injuries are more frequent than overuse shoulder injuries.
In conclusion, no shoulder rotators strength differences was found comparing rugby athletes with and without previous shoulder injuries.

REFERENCES

Table 2. Muscle isometric strength between dominant and non-dominant shoulders in control group and previous injured shoulder in injured group of rugby players. Data are presented as mean (standard deviation).

<table>
<thead>
<tr>
<th></th>
<th>Control group (n=16)</th>
<th>Injured group (n=15)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Dominant</td>
<td>Non dominant</td>
</tr>
<tr>
<td><strong>External rotation (f)</strong></td>
<td>13.75 (3.96)</td>
<td>10.56 (4.30)</td>
</tr>
<tr>
<td><strong>Internal rotation (f)</strong></td>
<td>14.41 (8.12)</td>
<td>14.49 (5.05)</td>
</tr>
</tbody>
</table>
INTRODUCTION
Fixation of prostheses to the humerus has traditionally been achieved by means of a cemented or uncemented stem inserted into the humeral canal. The uncemented stemmed implants are fixed by press-fit with the cortical bone. Recently, stemless implants have been developed that are fixed without cement to the humerus by press-fit with the trabecular bone. Such stemless implants are designed to allow an anatomical reconstruction of the humeral head, simplify the surgical technique and preserve as much bone stock as possible.

The long term success of uncemented stemless shoulder implants relies on osseointegration, which can only be achieved with adequate primary stability. The goal of this study was to evaluate the influence of various parameters on micromotion of a stemless humeral implant during various upper limb activities.

METHODS
A combined approach was taken. First, experimental tests on human cadaveric bones evaluated the influence of bone quality, implant sizing and load magnitude on the primary stability of a stemless implant. In a second step, a finite element (FE) model of one experimentally tested bone was created and validated by direct comparison of the simulation and experimental results. The FE model was then used to evaluate the implant micromotion on the whole implant surface, under a large set of physiologically relevant loading conditions.

Experimental tests
Eighteen cadaveric humeri free of visible deformities were used. The specimens were cut to keep approximately 10 cm of the proximal humerus and bone density was assessed from CT scans.

Sidus® Stem-Free Shoulder implants (Zimmer GmbH, Winterthur, Switzerland) were tested in this study. The anchors were implanted into the bone according to the surgical technique. The humerus was cemented in the specimen holder and fixed to a mechanical testing machine (Zwick 1456, Zwick Roell, Ulm, Germany). Load levels of 520 and 820N were applied to the humeral head and the resulting relative motion of the implant with the bone was measured using DVRT (SG-DVRT-8, Lord Microstrain, Cary, NC, USA, 2 micron resolution) and high resolution imaging (Prosilica GX1920, AVT, Stadtroda, Germany, 4.5 micron resolution). Micromotion was compared at 1, 25, 50 and 100 cycles and was seen to stabilize quickly with little variation between the first few cycles and after 100 cycles. Statistical significance of bone quality, load level and implant size on implant micromotion was tested using ANOVA.

Figure 1: Stemless humeral implant comprising a rough blasted titanium alloy anchor and cobalt chromium humeral head.

Figure 2: Experimental test setup showing the human humerus secured in the testing machine with the load...
applied to the implant humeral head. Micromotion of the implant is measured by means of a set of DVRT and a high resolution camera.

**Finite element model**

One of the experimentally tested humerus (right side of a 69 year old male donor) was selected for the creation of the FE model. The CT scan images of the intact humerus were segmented and 3D reconstruction was performed in Mimics (Version 14.1, Materialise, Leuven, Belgium). The humeral head resection height, version and inclination and implant position that had been measured in the experiment were reproduced in CAD (Unigraphics, Siemens PLM Software, Plano, TX). The assembly was exported to ANSYS (Workbench 15.0, Ansys Inc., Canonsburg, PA) for quadratic tetrahedral meshing and FE analysis. Average generic linear-elastic material properties were applied to the cancellous (210MPa and 0.4 Poisson’s ratio) and to the cortical bone (17GPa and 0.4 Poisson’s ratio) to match the average bone density of the tested bone. Coulomb frictional contacts were defined between the cancellous bone and the rough blasted faces of the anchor with a 0.6 friction coefficient [1].

The FE model was validated by simulating the configuration used in the physical test and direct comparison of the simulated micromotion values to those obtained experimentally with the high resolution imaging of the bone-implant interface. After validation of the FE model, physiologic glenohumeral joint resultant forces and moments representative of 29 different upper limb activities were applied to the FE model to evaluate the primary stability of the implant during in vivo use. In vivo force and moment components measured with instrumented shoulder implants were obtained from the publicly available orthoload database (www.orthoload.com) [2,3]. The 29 tested activities represented physiotherapy exercises, manual work or use of mobility aids (crutches and manual wheelchair).

**RESULTS**

The experimental tests revealed that bone density (P<0.0005) and load (P<0.0001) had a significant effect on implant micromotion. However the effect of implant size (P=0.123) was not statistically significant. Excellent agreement was found during FE model validation, with simulated and experimental micromotion values that were within 4μm.

The FE analysis demonstrated that the type of activity had a strong influence on micromotion. During all of the simulated upper limb activities, at least 99% of the implant surface experienced peak micromotion below 150 μm.

**Figure 3:** Micromotion distribution obtained from the FE simulation.

**DISCUSSION AND CONCLUSIONS**

This is the first study to investigate the in vitro and in silico micromotion behavior of a stemless humeral implant, looking at the influence of clinically relevant parameters. The experimental tests demonstrated the influence of bone quality and loading magnitude. The FE analysis showed that the variety of loading found during upper limb activities also plays a large role on the micromotion levels experienced by the prosthesis. However, peak micromotion below 150 μm, the generally accepted threshold below which bone ingrowth occurs, was measured over at least 99% of the implant surface during the entire set of simulated activities, even during very demanding activities that are not representative of the rehabilitation period. This study also demonstrates the need to include the variations of in vivo loading magnitude and orientation when testing micromotion in silico and/or in vitro.

**REFERENCES**

INTRODUCTION
The shape of the humeral head has been described as a sphere and this has provided basis for measuring the glenohumeral joint kinematics [1]. In the contrary, various studies have indicated that the native humeral head cannot be fully represented by a sphere [2]. Clinically, the importance of reproducing the normal anatomy of the proximal humerus in total shoulder arthroplasty (TSA) has been documented in literature [3]. The proximal humerus varies largely in shape and dimension and replicating its three-dimensional geometry will enhance the performance of the prosthesis [4]. Prosthetic component design is one of the major factors influencing the outcome of TSA surgeries since any change in the anatomy of the prosthesis may alter the shoulder biomechanics and consequently result in its failure [3]. Therefore, the focus of this study is to analyze the three-dimensional morphometric parameters of the proximal humerus of South Africans.

METHODS
3-Dimensional Reconstruction of the Proximal Humerus
Sixteen South African (SA) fresh cadaveric humeri and fourteen Swiss humeri were used in this study. The SA data consisted of 6 men and 2 women with ages ranging from 32 to 55 years while Swiss consisted of 3 men and 4 women with ages ranging from 19 to 77 years. The SA humeri were scanned using a CT scanner while the CT scans of the Swiss were obtained through a Swiss data repository. The Digital Imaging and Communications in Medicine (DICOM) files from the CT data were imported into a medical modelling software, MIMICS, for reconstruction. The 3D reconstructed model of the humerus (Figure 1) was imported as an STL file for further processing.

Reverse Modelling of the 3-Dimensional Surface Model
The STL data were generated as point cloud in a CAD software, SolidWorks. Thereafter, remodeling was done which involves defining the geometric features that is the detailed Referential Geometric Entities (RGEs).

Figure 1. Mimics Interface showing the 3D reconstruction of the proximal humerus

Identification of RGEs and Measurements
Anatomical reference points were defined as shown in Figure 2a and the parameters were measured based on the RGEs (Figure 2b).

RESULTS
Table 1 presents the summary of the measurements taken on the SA and Swiss proximal humeri. In table 2, a comparison is made between the SA (n=16), Swiss (n=14) and Caucasian (n=65) population data. The mean inclination angle for the SA proximal humeri (121.05º) was found to be lesser than that of the Caucasian (129.6º). On the other hand, the mean inclination angle for the Swiss (131.81º) was found to be greater than that of the Caucasian.
Table 1. Morphometric measurements of the SA and Swiss proximal humeri

<table>
<thead>
<tr>
<th>Parameters</th>
<th>SA (n=14)</th>
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<th>Switzerland (n=14)</th>
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<tr>
<td></td>
<td>Mean</td>
<td>Std Dev</td>
<td>Min</td>
<td>Max</td>
<td>Mean</td>
<td>Std Dev</td>
</tr>
<tr>
<td>Humeral head diameter (mm)</td>
<td>45.74</td>
<td>3.95</td>
<td>44.42</td>
<td>52.81</td>
<td>47.26</td>
<td>4.29</td>
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<tr>
<td>Articular surface diameter (mm)</td>
<td>42.42</td>
<td>3.77</td>
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<td>46.39</td>
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<td>Inclination Angle (degrees)</td>
<td>121.85</td>
<td>5.30</td>
<td>111.60</td>
<td>135.87</td>
<td>131.81</td>
<td>4.38</td>
</tr>
</tbody>
</table>

Table 2. Comparison of the morphometric measurements

<table>
<thead>
<tr>
<th>Parameters</th>
<th>SA Mean</th>
<th>Std Dev</th>
<th>Switzerland Mean</th>
<th>Std Dev</th>
<th>Caucasians Mean</th>
<th>Std Dev</th>
</tr>
</thead>
<tbody>
<tr>
<td>Humeral head diameter (mm)</td>
<td>45.74</td>
<td>3.59</td>
<td>47.26</td>
<td>4.35</td>
<td>46.2</td>
<td>5.4</td>
</tr>
<tr>
<td>Articular surface diameter (mm)</td>
<td>42.42</td>
<td>3.72</td>
<td>44.24</td>
<td>4.27</td>
<td>43.5</td>
<td>4.3</td>
</tr>
<tr>
<td>Inclination Angle (degrees)</td>
<td>121.08</td>
<td>5.30</td>
<td>131.81</td>
<td>4.30</td>
<td>129.6</td>
<td>2.9</td>
</tr>
</tbody>
</table>

DISCUSSION

In general, the SA proximal humeri were observed to be smaller than the Caucasian in all the parameters that were measured. However, the data measured showed that the Swiss proximal humeri are larger than the average Caucasian. The inclination angle was smaller which suggests that the SA proximal humeri are more inclined to the posterior than those of the Caucasian. It is interesting to note that the measured parameters on the Swiss proximal humeri are higher than the Caucasian. This explains the large variance which exists within population groups. Some of the existing shoulder prostheses have been designed based on the anthropometric data obtained from the Caucasian population [4, 5]. Therefore, the use of these imported shoulder implants in Africa and particularly in South Africa could result in mismatch in the SA patients. Although smaller prostheses could be introduced to the SA patients, other complications may arise. The glenoid may not be in full contact with the humeral head resulting in instability. Furthermore, there could be force imbalance on the glenoid eventually leading to glenoid loosening. The limitation of this study is the sample size used.

CONCLUSION

The morphometric data on the African shoulder is very limited and this study will significantly contribute to the shoulder data repository for the SA population. The morphometric parameters measured in this study will be useful in designing a South African shoulder prosthesis that mimics the native shoulder hence minimizing postsurgical complications.

REFERENCES


ACKNOWLEDGEMENTS

We acknowledge the support received from NRF Innovation Fellowship and Schlumberger Foundation Faculty for the Future through a Post-Doctoral Fellowship awarded to one of the authors.
Comparison between an anatomic vs. reverse shoulder prosthesis by means of an artificial shoulder simulator

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IMES Institute for Mechanical Systems, ZHAW Winterthur, Switzerland
Smith & Nephew Orthopedics AG, Aarau, Switzerland

INTRODUCTION
Orthopedic implants for the shoulder may still be optimized to improve their functional behavior. In order to acquire more knowledge about the acting muscle and joint forces and load distribution, a comprehensive analysis with an anatomical and a reverse implant was carried out. Previous studies focused on muscle forces measured during static equilibrium in a fixed abducted position of the arm [1]. The present study performed measurements during dynamic abduction. Prior assumptions predict a decrease of necessary muscle forces for the reverse implant [1] due to a more advantageous pivot point compared to the anatomical implant. It is also presumed that the shear force is increased for the reverse implant compared to the anatomical implant [1]. The aim of the study was to investigate shearing and compressive joint reaction and muscle forces for the reverse and anatomic prosthesis designs.

METHODS
Shoulder Simulator
All tests were performed by using an experimental shoulder simulator [2] and artificial humeri (Synbone). Artificial tendons (latchets) were attached on tendon insertion areas on the humeri representing the rotator cuff (Figure 2) and tied to bowden cables of the simulator (tendons). The muscles (M. Deltoideus, M. Supraspinatus, T. Minor, M. Infraspinatus, M. latissimus and M. Pectoralis) were simulated by DC-motors and nylon cords and fixed to the latches.

Prosthesis preparation
The proximal part of the humerus was cut off and a respective implant (Smith&Nephew) was inserted into the shaft of the humerus. Additionally a glenoid made of polyethylene (Mathys AG Bettlach) was used for the anatomical implant. The inverse implant itself consists of a concave socket inserted into the humerus and a metallic head fixed to the scapula instead of the glenoid cavity.

Tests
Three different settings have been defined for the testing: Setting 1: An anatomic implant (fig. 2) using 100% DELT without SSP (DELT). Setting 2: Combination of DELT and SSP in a ratio of 2:1 (DELT-SSP). Setting 3: A reverse implant with the deltoideus as abductor (REV) (fig. 2) without SSP. Each test consisted of five abduction cycles from 0° to 65° including scapular rotation (angular ratio 3:1 humerus - scapula). From the data of each of the tests joint- and muscle force values for distinct angles (15°, 30°, 45° and 60°) were extracted, averaged and p-values determined. The weight of the arm was 3.5 kg whereby the humerus length was fixed to 30cm (distance from the humeral head center to the lower rim of the attached weight).

Figure 1: Shoulder simulator with an integrated metallic upper arm dummy.
Figure 2: Examples of the anatomical shoulder implant (left) and reverse shoulder implant (right).
RESULTS
Results of the muscle forces of all three settings are shown in fig. 3. Total glenohumeral forces are presented in fig. 4.

Figure 3: Comparison of muscle deltoid force for different prosthesis configurations. Statistical differences based on 95% confidence interval (error bars) indicated with asterisks.

Figure 4: Comparison of glenoid force for all three settings: DELT, DELT-SSP, REV. Statistical differences based on 95% confidence interval (error bars) indicated with asterisks. Legend indicated in fig. 2.

The Deltoideus force in REV was higher for abduction angles smaller than 30° compared to those in DELT. However compared to DELT-SSP the consequent force was increased up to 60°.

The shear force (Fx) in REV was highly increased (>2 fold) for short angles (<30°) compared to DELT whereas compared to DELT-SSP the shear force was converging for increasing angles.

The compression force (Fz) in REV was increased for all considered angles compared to DELT. For lower angles (<30°) the compression force was consistent for DELT and DELT-SSP. In case of DELT-SSP the compression force was significantly higher than for REV for angles ≥30°.

DISCUSSION AND CONCLUSIONS
The reverse implant requires almost double of the Deltoideus force up to 30° compared to DELT-SSP, which may be explained by the supporting effect of the intact SSP in the anatomic prosthesis. For higher abduction angles (>30°), the deltoid force was still increased. In DELT-SSP the SSP increased compressive force but decreased shear force compared to DELT as already confirmed in literature [1]. Increased shear force was also observed for REV up to 45°, while compressive force was increased from 30° to 60°, meaning that the reverse implant increases resulting joint reaction force compared to DELT. Compared to DELT-SSP the shear force was strongly increased for all considered angles but compressive force was decreased as described in literature [3]. The reverse implant seems to require higher Deltoideus force which might lead to a higher risk of abrasion in the glenohumeral joint. These findings did not confirm the prior assumption of a reduced Deltoideus force in the reverse prosthesis, because it strongly depends on the current abduction angle. The results have to be confirmed by a higher number of tests, in particular including the high anatomical variability of acromion lateralization or glenoid inclination. These parameters and their influence in shoulder function are recently discussed, describing the so called Critical Shoulder Angle CSA [4].

REFERENCES
2. Baumgartner, Med Biol Eng Comp, 52:293-9, 14

ACKNOWLEDGEMENTS
This study was financially supported by the Swiss Science Foundation SNF Grant No. 320030_147110.
INTRODUCTION
Longevity of Total Shoulder Arthroplasty necessitates a well fixed glenoid. Historically, the dominant failure mode for the traditional cemented shoulder replacement has been due to fixation failure [1,2]. For a reversed configuration shoulder, where the loading could be considered more challenging for the fixation, success has been found in the use of metallic (often titanium alloy) baseplates held in position by two or more peripheral screws. Previous work by the author has demonstrated that an optimisation of screw angulation, thickness and length all contribute to enhance primary stability of the baseplate [3], demonstrating the value-added from poly-axial screws. However, scapular anatomy does not permit the surgeon to blindly align or size the screws without risking damaging the surrounding soft tissues if the screw would perforate the bone. In order to optimise the fixation, some companies have developed pre-op planning techniques for the surgeon. Our hypothesis is that a guided approach to the positioning of the peripheral screws (insertion site, angulation) would yield the potential for optimising the fixation of the glenoid.

METHODS
3D models of 24 scapulae (7 female, 12 male, 5 unknown gender) were reconstructed from CT data using the software package AVIZO (Mercury Computer Systems). All scapulae were orientated into a common coordinate frame (Figure 1). A sensitivity analysis was performed to determine the influence of small misalignments of the scapulae, in terms of their effect on the predictive power of the analyses, and this was found to be negligible. An oval glenoid implant template (34x42mm) was centred over the shallowest point on the glenoid and aligned with respect to the scapular axis. The dimensions of this template were selected to accommodate the full range of glenoid dimensions considered in this study. This template comprised a grid of 200 potential insertion sites for any peripheral screws on the glenoid surface, divided into four zones of interest depending on location. These ‘zones’ are defined as anterior (45 insertion sites), posterior (45 sites), superior (55 sites) and inferior (55 sites). From each screw insertion site, the screw was ‘grown’ along a range of possible screw trajectories until a contact is detected against the inner surface of the bone. Screws were projected in an arc of 90 degrees according to the insertion zone (i.e. a superior screw would only be aligned in a superior direction up to 20 degrees from the scapular axis), giving a total of 57 directions per insertion site. In total, 11,400 potential screw projections for either a 3 or 5mm diameter screw were tested per specimen.

RESULTS
The results of the analyses were compiled and presented in Tables 1 and 2. When comparing between the different insertion zones, peripheral superior or inferior screws could be positioned much deeper than anterior or posterior screws. In fact, AP peripheral screws were frequently shorter than 20mm, the limits of commercially available screw lengths.

<table>
<thead>
<tr>
<th></th>
<th>S (mm)</th>
<th>I (mm)</th>
<th>A (mm)</th>
<th>P (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>“Blind” screw alignment</td>
<td>16.2mm (9.6-26)</td>
<td>22.6mm (11.9-29.2)</td>
<td>9.8mm (4-20.8)</td>
<td>6mm (3.8-8.5)</td>
</tr>
<tr>
<td>“Navigated” screw alignment</td>
<td>60.1mm (26-97)</td>
<td>46.6mm (22.8-85.3)</td>
<td>24.1mm (13.1-48.2)</td>
<td>11.1mm (7.4-20.1)</td>
</tr>
<tr>
<td>% gain (Navigated vs Blind)</td>
<td>389% (208-729)</td>
<td>207% (159-377)</td>
<td>272% (160-545)</td>
<td>187% (148-341)</td>
</tr>
</tbody>
</table>

Table 1: Results of screw depth analysis for all 24 bones for 3mm diameter screws
### Table 2: Results of screw depth analysis for all 24 bones for 5mm diameter screws

<table>
<thead>
<tr>
<th>5mm</th>
<th>S</th>
<th>I</th>
<th>A</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>“Blind” screw alignment</strong></td>
<td>7.9mm (12.7-18.1)</td>
<td>15.8mm (9.9-22.8)</td>
<td>7.3mm (2.9-17.4)</td>
<td>4.9mm (1.7-7.7)</td>
</tr>
<tr>
<td><strong>“Navigated” screw alignment</strong></td>
<td>39.3mm (21.5-88.9)</td>
<td>35.5mm (21.7-47.5)</td>
<td>18mm (6-34.8)</td>
<td>11.8mm (7.5-39.9)</td>
</tr>
<tr>
<td>% gain (Navigated vs Blind)</td>
<td>312% (200-514)</td>
<td>231% (181-342)</td>
<td>273% (133-684)</td>
<td>258% (157-594)</td>
</tr>
</tbody>
</table>

With regard to the potential percentage gain from navigated versus blind screw insertion, on average across the 24 specimens tested herein, an improvement of screw length of 389% (39.9mm) was possible for a superior screw with 3mm diameter and 312% (31.4mm) for a 5mm diameter. An inferior screw length could be improved by between 207% (24mm) for a 3mm screw and 231% (19.7mm) for a 5mm screw. Length gains for AP screws due to navigation or pre-op planning were limited by the available bone stock in these zones.

**DISCUSSION AND CONCLUSIONS**
Pre-operative planning to allow a surgeon to maximise the depth of peripheral screws has been clearly demonstrated to increase screw length versus blind and non-assisted insertion. This would in turn facilitate improved longevity for metal backed glenoids.

**REFERENCES**
Pre-Operative Planning and Accurate Implantation Can Increase Impingement Free Range of Motion in Reverse Shoulder Arthroplasty – A Cadaveric Validation

Andreas Kontaxis, Julien Berhouet, Daniel Choi, Xiang Chen, David Dines, Russell Warren, Lawrence Gulotta

INTRODUCTION
Reverse Shoulder Arthroplasty (RSA) is a popular treatment for patients with cuff tear arthropathy [1], humeral fractures, or other challenging arthropathies [5]. Clinical and biomechanical studies have shown the benefits of RSA [1,2] in pain relief and functional recovery, but concerns still remain regarding impingement, range of motion (ROM) and scapular notching [3].

Recent studies have highlighted the importance of accurate prosthesis placement [3] and reported the use of patient-specific guides in aim to improve accuracy in glenoid component positioning for RSA. Most of the guides that have been assessed in the literature have shown promising results on achieving accuracy. However there are no studies to demonstrate how accurate placement can benefit RSA. The aim of this investigation was to investigate whether impingement free ROM can be improved when a pre-operative plan and accurate implantation is used compared to the traditional RSA surgeries.

METHODS
Types of Surgery: This study performed RSA surgery in 20 cadaveric specimens with two methodologies
i) ‘Traditional’ surgery where glenoid baseplate placement was determined with the help of standard surgical instrumentation
ii) ‘Guided’ surgery where subject specific tools were used to aid the surgeon placing the glenoid component according to a pre-operative plan.

Surgeries were performed from a single surgeon with experience in RSA. Cadavers and order of the surgery were randomized. A single size commercial RSA prosthesis was used (Biomet Comprehensive, Biomet, Warsaw, Indiana, size: 28mm diameter glenoid baseplate; 36mm diameter standard glenosphere without offset; size 4 humeral stem).

For the ‘Traditional’ method the manufacturer’s instrumentation set was used to place the steinman pin on the glenoid. For the ‘Guided’ surgeries, the position was defined according to a pre-operative plan.

Pre-Operative plan: Virtual custom models were created for each scapula from pre-operative CT-scans. A virtual RSA surgery was performed in each model where the glenoid component was placed in an initial position that followed surgical recommendations. Using a contact detection algorithm the impingement free range of motion was calculated for some standardised motions: i)Abduction, ii)Scaption, iii) Forward Flexion. An iterative glenoid placements were tested to define the glenoid position that maximized the impingement free ROM.

Subject specific guides: Specific guides were manufactured by a 3D printer following published recommendations (4). The objective of the guide was to place the Steinman pin into the position that was defined by the pre-operative plan. The design of the guides used subject specific anatomical landmarks of the anterior glenoid and coracoid foot to aid the surgeon place the steinman pin (Figure 1.a).

RESULTS
The majority of the cadavers had no osteoarthritis (there were only 4 specimens with limited presence...
of osteophytes). The average age and BMI of the specimens in the ‘Traditional’ method were 31.4 (SD 7.0) and 71.0 yo (SD 8.0) respectively and for the ‘Guided’ method were 31.0 (SD 7.2) and 71.4 yo (7.0).

The results of the glenoid placement showed a deviation from target for the ‘Guided’ specimen (compared to pre-op plan) for version (Table 1).

<table>
<thead>
<tr>
<th></th>
<th>Guided</th>
<th>Traditional</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pre-Op</td>
<td>Post-Op</td>
<td>Pre-Op Plan</td>
</tr>
<tr>
<td>Version</td>
<td>4.2(4.7)</td>
<td>5.8(7.7)</td>
</tr>
<tr>
<td>Tilt</td>
<td>8.2(2.8)</td>
<td>9.6(4.2)</td>
</tr>
</tbody>
</table>

Table 1: Pre and Post-Operative glenoid version and tilt for both types of surgeries. Results show the average version and tilt in degrees (SD)

The guided surgery also resulted in a glenoid placement that was on average 4.2 mm (SD 4.0) more inferiorly compared to the ‘Traditional’ method cadavers (p<0.05).

This difference in baseplate placement had also an effect on the impingement results. The average inferior impingement was recorded at 10.4° (SD 4.3) and superior impingement averaged at 67.9° (SD 4.3) for the guide surgery specimens. Even if the results were different from the pre-operative plan, it was better than the cadavers that received surgery with the standard instrumentation tools (Traditional surgery: average inferior impingement: 29.6°, SD 13.7, average superior impingement 59.3°, SD 1.54).

DISCUSSION AND CONCLUSIONS
The results showed that the subject specific guides were fairly accurate following a pre-operative plan, but they deviated from target more for version than tilt. Comparing the version and tilt results between ‘Guided’ and ‘Traditional’ surgeries, there was not a significant difference. This was expected since the cadaveric specimens had no arthritis or abnormal glenoid deformities. Previous studies showed that it is very challenging for surgeons to successfully correct glenoid version and tilt on highly eroded glenoids intra-operatively.

The ‘Guided’ surgeries also showed a more inferior glenoid placement compared to the ‘Traditional’ surgeries. The latter translated in a better functional outcome with a larger free impingement ROM justifying the use of subject specific guides. However the results were not as good as the pre-operative plan/model had predicted. This was probably because of the deviation from target that was observed (in glenoid placement) and also due to soft-tissue tensioning (deltoid muscles) that was not been accounted in the pre-operative plan/model.

REFERENCES
INTRODUCTION
Previous studies examine the acute effect of a greater thoracic kyphosis on scapular posture comparing changes in scapular positions when subject switches between the “normal/ideal” trunk posture and the adoption of a slouched posture [1, 2]. However, no information exists about the effect of a smaller thoracic curvature (“flat thoracic spine”) on the scapular motion during arm elevation. Moreover, it is unknown what the changes on scapular motion pattern are when the “flat thoracic spine” is a result of a chronic morphological adaptation such as in classical ballet dancers. Indeed, a flat thoracic configuration and a backward shoulder posture (“non-winged scapulas”) are often associated with ballet dancers. Thus, the purpose of the present study was to compare scapular posture associated with three arm elevation positions in individuals with a smaller (flat) and greater thoracic spine curvature. We hypothesized that individuals with less thoracic kyphosis would display more scapular upward rotation as well as greater external rotation and posterior tilting.

METHODS
In order to create distinct groups based on the magnitude of thoracic kyphosis curvature a postural analysis was performed over a large population of potential participants. All participants signed an informed consent form approved by the university ethics committee. Thoracic spine curvature on the sagittal plane (i.e. kyphosis) was measured on resting standing position by mean a Flexible Curve (Faber-Castell, Germany) which is a malleable band of metal covered with plastic and approximately 60 cm in length. The ruler can be bent in only one plane and retains the shape to which it is bent. The subject was instructed to stand up straight and as tall as possible, and the Flexible Curve ruler was aligned to the anterior-posterior curves of the spine from C7 to T12. The ruler was then placed flat on graph paper (mm) and its outline was traced. A straight line was then drawn from the ruler position of C7 to T12 that corresponded to the length of thoracic kyphosis (length). The height of the thoracic kyphosis (height) was determined by drawing a perpendicular line from the highest point in the thoracic curve to the point at which it intersected the straight line drawn from C7 to T12. A percentage ratio between the height and length of the thoracic kyphosis, the kyphosis index (KI) was calculated according to Barret et al. [3] by applying the formula (height / length) x 100. The higher scores of the KI indicate greater degrees of thoracic kyphosis. Based on these measures a cut-off value of 8.5 for the KI was used as postural criteria to distinct 2 groups of participants: the “Flat Thoracic Spine” (FTS) group (N=15; Age = 19.3 ± 4.3 years; Height = 167.6 ± 8.6 cm; Mass = 47.6 ± 8.1 Kg) and the “Kyphotic Thoracic Spine” (KTS) group (N=15; Age = 19.6 ± 3.6 years; Height = 168.1 ± 6.8 cm; Mass = 59.6 ± 4.8 Kg). The FTS participants were recruited among a population of classical ballet dancers with an average practice time of 12.7 ± 2.4 years. A 6DOF electromagnetic tracking device (Hardware: “Flock of Birds system” Ascension Technology; Software: Motion Monitor v 7.0) was used to record humeral and scapular 3D posture in 3 arm positions: rest position (± 15º elevation), 90º abduction and maximum arm elevation angle. A set-up of 4 sensors (100 Hz) was used: the thorax sensor placed over the spinous process of T1; the scapula sensor placed over the broad flat surface of the posterolateral acromion of the dominant scapula; the humeral sensor placed over the posterior aspect of the dominant humerus, distal to the triceps muscle belly, by mean of a customized cuff. A 4th sensor mounted on a hand-held stylus (±6.5cm) was used for bony landmarks digitalization in order to link sensors to local anatomical coordinate systems (LCS) and subsequently calculated segments and joint rotations by combining the LCSs with the sensor motions. Segments LCSs and joint rotations definition were made according to the shoulder ISB standardization protocol [4]. The kyphosis index and the scapular rotations were analyzed using standard tests for normality (Shapiro-Wilk Test) and were found to satisfy assumptions of normality [5].
An independent-samples t-test was run to examine differences in kyphosis index as well as on scapular rotations in each of the three arm positions. The relationship between the kyphosis index and each of the scapular rotations was examined using the Person product-moment correlation coefficients. For all statistical tests, specific software (SPSS Statistics 19.0) was used and critical level of \( p < .05 \) was considered statistically significant for all analyzes.

RESULTS

Differences between both groups of subjects with respect to the thoracic spine curvature were confirm by statistical analysis \( t(28) = 10.9; p = 0.0; \) Mean difference: 1.9; 95CI: 2.3 – 1.6] with the FTS group displayed smaller kyphosis index (KI) [FTS group Mean (SEM): 7.7 (0.7); KTS group: 9.7 (1.2)]. Significantly differences (\( p<0.5 \)) were found between groups of subjects on all scapular rotations and arm positions. In all arm positions (rest, 90º and maximum) the FTS group (dancers) displayed significantly greater scapular upward rotation angles and smaller internal rotation angles. In this group, scapula was more posterior tilting, at 90º, and more anterior tilting at rest and maximum arm elevation (Table 1).

Table 1: Scapular rotation angles [Mean (SEM)].

<table>
<thead>
<tr>
<th>Groups of Participants</th>
<th>M.D. 95%C.I.</th>
</tr>
</thead>
<tbody>
<tr>
<td>FTS (Dancers)</td>
<td>KTS (Non-Dancers)</td>
</tr>
<tr>
<td>Sx</td>
<td>1.5 (1.2)</td>
</tr>
<tr>
<td>Sy</td>
<td>35.4 (1.5)</td>
</tr>
<tr>
<td>Sz</td>
<td>14.7 (1.2)</td>
</tr>
<tr>
<td>REST</td>
<td></td>
</tr>
<tr>
<td>Sx</td>
<td>19.7 (1.3)</td>
</tr>
<tr>
<td>Sy</td>
<td>40.5 (1.3)</td>
</tr>
<tr>
<td>Sz</td>
<td>6.9 (1.4)</td>
</tr>
<tr>
<td>90º</td>
<td></td>
</tr>
<tr>
<td>Sx</td>
<td>45.3 (3.1)</td>
</tr>
<tr>
<td>Sy</td>
<td>43.0 (1.7)</td>
</tr>
<tr>
<td>Sz</td>
<td>3.5 (1.6)</td>
</tr>
</tbody>
</table>

A significantly negative correlation was found between KI and upward rotation and spinal tilt in all experimental condition with an exception for upward rotation at maximum elevation angle. A significantly positive correlation was found for scapular protraction at rest and 90º arm positions.

DISCUSSION AND CONCLUSIONS

Our results are consistent with previous reports examining the effects of the magnitude of thoracic curvature on 3D scapular posture. Kebaetse et al. [2] found that the increased thoracic kyphosis placed the scapula in a resting position with greater medial rotation, upward rotation, and anterior tip as compared with an upright posture. Finley et al. [1] found that when an increased thoracic kyphosis (sloached posture) was adopted, there were significant decreases in the posterior tip and lateral rotation of the scapula, but there was no significant change in the magnitude of the upward rotation of the scapula. More recently Thigpen et al. [6] found in individuals with forward head and rounded shoulder a greater scapular internal rotation with less serratus anterior activity during two overhead tasks (loaded arms flexion and forward overhead reaching task) as well as greater scapular upward rotation and anterior tilting, during the arm flexion task.

A significant relationship was found between thoracic spine curvature and scapular posture, particularly at rest and 90º arm elevation, such that smaller KI values are associated with an upward and anterior tilt scapular position.

In conclusion, the results of this study reveal that females with flat thoracic spine displayed greater upward, external rotation angles and anteriorly tilted in all arm positions (rest, 90º and maximum), with the exception at 90º, where scapula assume a posterior tilting position.

REFERENCES

Is it possible to help prevent scapula injuries in baseball pitchers?

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2INAIL Prosthetic Center, Vigorso di Budrio, Italy
3Orthopedic Clinic University of Parma, Parma, Italy
4Sport Medicine Unit, Loyola University Hospital, Chicago, US

INTRODUCTION
Injury in the throwing shoulder of baseball pitchers and overhead throwing athletes is common at the collegiate and professional level. During overhead throwing the scapula on one hand acts as a stable base of support between humerus and trunk, and on the other allows for wide ranges of motion (ROM) at the gleno-humeral joint. This ability to perform wide shoulder movements has been associated to a potential source of shoulder injury [1]. The posterior rotator cuff may become eccentrically fatigued in the throwing motion. Over time, this may lead to a thicker posterior capsule. The mechanism causing a hypertrophied posterior capsule is theorized to be a physiologic healing response and can change the physiological shoulder biomechanics [2]. Previous studies assessed scapular kinematics using electromagnetic tracking devices, demonstrating that patients with impingement have a reproducible pattern of shoulder dyskinesia [1]. In addition, they showed a correlation between posterior shoulder tightness and forward scapular posture and confirmed that baseball players have more forward scapular posture of the dominant arm [3]. Capsular stretching programs might help restoring a normal scapulo-humeral coordination, helping in injury prevention. The aim of this study is to evaluate the scapular kinematics of asymptomatic baseball pitchers, before and after a four-weeks posterior stretching protocol.

METHODS
Subjects and stretching protocol
Eleven asymptomatic collegiate baseball pitchers were involved in the study. Pitchers were randomly separated in two groups: group A (6 pitchers) underwent four weeks of a regimented therapy protocol; group B (5 pitchers) did not received any treatment. The stretching protocol consists of simple posterior capsular stretching exercises performed by the team physiotherapist once a week (prone stretching, supine stretching with internal rotation, supine stretching), and also home exercises (sleeper position, adduction/cross body stretching, and terminal shoulder extension exercises using a towel), performed by athletes twice a week.

Measurement protocol
All pitchers from both groups (A and B) were tested on two separate days: at the first day of the study (S1) and after four weeks (S2). The ISEO protocol (Fig. 1) was used to collect the kinematics of trunk, scapula and humerus [4] of both the throwing side (TS) and the contralateral side (CS), by means of inertial and magnetic sensors. The Scapulo-Humeral Coordination (SHC) during humerus elevation in the sagittal (FlexExtension - FE) and scapular (AbAdduction - AA) plane was collected: three repetitions for each movement. All scapula rotations were considered: protraction-retraction (PRRE); medio-lateral rotation (MELA); posterior-anterior tilting (PA). SHC was analysed by means of coordination plots where the kinematic angles of humerus (X-axis) and scapula (Y-axis) are reported.

Data analysis
For each subject and coordination plot, the mean SHC pattern among the three repetitions was computed; a common humeral ROM (from 10° to 110°) was considered and divided in steps of 5°.
In order to verify if biological modifications occurred, firstly SHC of TS and CS were compared to age stratified reference bands (RB), computed from asymptomatic subjects [5]. Secondly, the scapula ROM (sROM) variations of TS between S1 and S2 were computed and compared to the Smallest Detectable Difference (SDD) of ISEO [6]:

- If $|\Delta \text{ROM}| > \text{SDD}_{95\%}$, the difference between sROMs is due to biological variations;
- If $|\Delta \text{ROM}| < \text{SDD}_{95\%}$, the difference between sROMs is due to the reapplication of the protocol.

**RESULTS**

The comparison of the SHC patterns with RBs showed that both TS (in S1 and S2) and CS of all subjects are within the correspondent reference bands, except one subject from group B that presented alterations at TS only in S1. Figure 2 shows an example of SHC patterns VS RB comparison for the scapula MELA during FE.

From the comparison of $\Delta \text{ROM}$ with SDD values, group B showed that significant variations occurred only for one subject in MELA ROM during FE and AA and PA during AA. On the contrary, 4 out of 6 pitchers that received the therapy showed clear signed of SHC alterations. Specifically, 4 subjects (67%) had variations in MELA during FE and 3 (50%) in MELA during AA, and 1 subject in PA during both FE and AA. An example of $\Delta \text{ROM}$ analysis is reported in Table 1.

These kinematic changes were all towards the RB mean patterns, showing an improvement of the SHCs of TSs after the 4-weeks physical therapy.

**DISCUSSION AND CONCLUSIONS**

This study highlights how posterior capsular stretching protocol can favor the maintenance of the SHC in asymptomatic baseball pitchers. All pitchers of group A underwent the stretching protocol even if in S1 they had not presented an altered SHC compared to reference bands. The results demonstrate that there are statistically significant differences in the kinematics of several athletes from the “treated group” (group A). It is also important to notice that variations in group A occurred for both movements (FE and AA), strengthening the conclusion that the variation was real.

The results of the study can indicate that, in order to prevent the pathologic cascade linked to these sports activities, this physical training protocol might become integral part of the normal daily exercises of baseball pitchers and over-head athletes.

**REFERENCES**

INTRODUCTION

Overhead athletes are predisposed to shoulder injuries due to the repetitive movement, high velocities and large ranges involved in throwing. Several factors have been associated with shoulder injuries in overhead athletes, such as alterations in muscle balance and rotation range of motion, sensorimotor deficits and muscle fatigue. Although previous studies have shown that muscle fatigue results in scapular kinematics alterations in healthy subjects, studies on overhead athletes have found minimal alterations (1-3). However, these studies have assessed scapular kinematics during constrained movements, such as arm elevation and diagonal patterns. Since scapular kinematics differ between constrained and functional arm movements (4), we hypothesized that assessing throwing motions would be more appropriate to detect possible changes in overhead athletes following fatigue. Therefore, the aim of this study was to investigate the effects of muscle fatigue on scapular kinematics of overhead athletes during the throwing movement. Considering that the late-cocking phase, when the shoulder reaches the greatest external rotation (ER), is the most critical moment during throwing (5), scapular kinematics was analysed at this position.

METHODS

Subjects
Thirty healthy overhead athletes (21 male and 9 female, mean age 24±12.8 years), with a mean of 5.4 ± 3.8 years of practice and participation in university-level competition, were assessed. This study was approved by the Ethics Committee of the University.

Procedures
The Flock of Birds electromagnetic tracking system integrated with the Motion Monitor software was used to capture three-dimensional kinematics, at a sampling rate of 100 Hz. Receivers were attached to the sternum, acromion and distal humerus. Anatomical coordinate systems were determined following the International Society of Biomechanics recommendations (6). Surface EMG was collected with a 8-channel Bagnoli EMG System (DelSys, Boston, USA), using active, double-differential electrodes, at a sampling frequency of 2000 Hz. Electrodes were positioned over the serratus anterior, upper and lower trapezius, and maximal isometric voluntary contractions (MVIC) were performed prior to data collection for normalization of EMG amplitude. Data were collected pre and post fatigue during three trials of a simulated, low-velocity throwing motion, as previously described (7). A force sensing resistor sensor (Biometrics Ltd) was used to determine the ball release moment. The fatigue protocol consisted of quickly throwing a rubber ball against a wall, while seated on a chair, until reaching 8 points or more in the Borg’s Rate of Perceived Exertion Scale (1). For EMG fatigue quantification, subjects performed a submaximal isometric contraction at 15% MVIC of flexion, immediately before and after the fatigue protocol.

Data analyses
Data were processed using a program written in Matlab. Median power frequency (MPF) was calculated from the raw EMG signal collected during the submaximal isometric contractions. A decrease of 8% in MPF was considered indicative of muscle fatigue (8). Scapular and humeral positions relative to the thorax were extracted at the maximal humeral ER position. The EMG signal was filtered with a bandwidth from 30 to 450 Hz and the root-mean-square (RMS) was calculated using a 20-ms moving window with 50% overlap. Mean RMS was calculated for two movement phases: arm cocking (from the initiation of the movement to maximal humeral ER) and acceleration (from maximal humeral ER to ball release).

Statistical analysis was performed with SPSS 22. Pearson correlation was used to investigate possible relationships between changes in maximal humeral ER and scapular position changes. Each scapular and humeral position was compared pre and post fatigue.
using paired t-tests. EMG amplitude was analyzed using a 2-way repeated-measures ANOVA, with throwing phase and fatigue as within-subject factors. Significance level was set at 5%. Cohen's d effect size was calculated for significant differences.

RESULTS

Pearson correlation showed no association between humeral ER and scapular kinematic changes (r-values ranged from -0.22 to 0.21, p>0.05). There was a significant decrease in humeral ER following fatigue (effect size=0.23), but no significant change in scapular rotations, humeral elevation and horizontal abduction (Table 1). Serratus anterior presented a MPF decline higher than 8% in 77% of the subjects and RMS presented a significant main effects for throwing phase (effect size=0.80) and fatigue (effect size=0.38) (Table 2).

Table 1. Kinematics at maximal humeral external rotation of throwing motion, before and following fatigue protocol.

<table>
<thead>
<tr>
<th></th>
<th>Pre-fatigue</th>
<th>Post-fatigue</th>
<th>P value</th>
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</thead>
<tbody>
<tr>
<td>Humerothoracic</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>External rotation</td>
<td>74.9±10.6</td>
<td>72.5±10.8</td>
<td>0.02*</td>
</tr>
<tr>
<td>Horizontal abduction</td>
<td>35.3±12.7</td>
<td>34.6±17.7</td>
<td>0.78</td>
</tr>
<tr>
<td>Elevation</td>
<td>86.2±12</td>
<td>84.6±14.3</td>
<td>0.39</td>
</tr>
<tr>
<td>Scapulothoracic</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Upward rotation</td>
<td>20.9±6.7</td>
<td>21.07±8.8</td>
<td>0.88</td>
</tr>
<tr>
<td>Internal rotation</td>
<td>22.6±9.9</td>
<td>23.2±10.8</td>
<td>0.69</td>
</tr>
<tr>
<td>Tilting</td>
<td>12.1±6.7</td>
<td>10.6±7.6</td>
<td>0.09</td>
</tr>
</tbody>
</table>

Data are mean ± standard deviation, presented in degrees. *Significant difference following fatigue.

Table 2. Mean RMS (% MVIC) at each throwing phase, before and following the throwing fatigue protocol.

<table>
<thead>
<tr>
<th></th>
<th>Arm cocking</th>
<th>Acceleration</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Pre-fatigue</td>
<td>Post-fatigue</td>
</tr>
<tr>
<td>Upper trapezius</td>
<td>14.5±1.7</td>
<td>16.5±1.9</td>
</tr>
<tr>
<td>Lower Trapezius</td>
<td>12.9±1.1</td>
<td>14.9±1.3</td>
</tr>
<tr>
<td>Serratus Anterior</td>
<td>19±2.1</td>
<td>22.8±2.8</td>
</tr>
</tbody>
</table>

*Main effect for fatigue (p=0.03) and phase (p=0.006).

DISCUSSION AND CONCLUSIONS

This study shows that muscle fatigue does not alter scapular kinematics of healthy overhead athletes during the late-cocking phase in a simulated throwing task. These results are in accordance with previous studies that have found only small alterations on scapular kinematics during constrained movements (1-3). These findings reinforce a previous suggestion that healthy athletes may present some adaptive mechanism which might prevent alterations in scapular movements, despite the presence of serratus anterior fatigue (1). Increased EMG amplitude of the serratus anterior following the fatigue protocol may be an indicative of myoelectric fatigue, but may also have contributed to the maintenance of scapular position during late-cocking. Regarding humerothoracic kinematics, muscle fatigue had only a small effect on ER. Our findings suggest that scapular alterations caused by muscle fatigue are probably not the dominant mechanism of shoulder injuries in overhead athletes. However, muscle fatigue should not be discarded as a risk factor for shoulder injuries in overhead athletes, since it may cause other detrimental alterations, such as proprioceptive deficits.

REFERENCES


ACKNOWLEDGEMENTS

Gisele G. Zanca was recipient of research fellowship from São Paulo Research Foundation (FAPESP) (#2012/05146-0).
INTRODUCTION

Posterior capsule tightness (PCT) is a prevalent soft tissue alteration in those who perform overhead sports or activities1,2 and is frequently associated with shoulder pain.3-6 Increased internal rotation and anterior tilt of the scapula, and increased anterior and superior humeral translations were already identified in asymptomatic throwers athletes, asymptomatic non-throwers individuals with PCT7-10 and individuals with shoulder pain.11-14 However, it has not yet been determined if the same alterations are present in subjects with PCT and shoulder pain. Studies analyzing the interaction of pain and PCT are important to better understand the real contribution of the tightness to shoulder pain. Therefore, the purpose of this study was to assess scapular and humeral motions in individuals with PCT with and without shoulder pain.

METHOD

Eighty individuals participated in this study and were divided in 4 groups: asymptomatic to shoulder pain without PCT (13 women and 7 men; 28.70±7.10 years; 67.55±12.30 kg; 1.69±0.09 m); asymptomatic to shoulder pain with PCT (13 women and 7 men; 28.45±7.33 years; 66.34±10.89 kg; 1.67±0.10 m); symptomatic to shoulder pain with PCT (12 women and 8 men; 33.05±9.28 years; 69.85±11.12 kg; 1.68±0.11 m); and symptomatic to shoulder pain without PCT (11 women and 9 men; 28.10±8.30 years; 69.72±11.92 kg; 1.69±0.12 m). The PCT was determined by the low flexion test as described by Borstad and Dashottar.15 This test was performed with the individuals in the standing position. They were passively positioned in 60° of arm elevation in the sagittal plane, and then the evaluator measured the bilateral internal rotation range of motion with a digital inclinometer aligned with the radial styloid. A difference of at least 7° in internal rotation between arms was used to classify individuals with tightness. Flock of Birds® electromagnetic tracking system was used to evaluate scapular kinematics and humeral head translations during elevation of the arm in the sagittal plane. Local coordinate systems were established for the trunk, scapula, and humerus using the digitized landmarks following the protocol recommended by the International Society of Biomechanics.16 The YX’Z” Euler sequence16 was used to describe scapular motions relative to the trunk in the sequence of internal/external rotation, upward/downward rotation and anterior/posterior tilt. The helical axis was used to determine humeral translations.12 Three repetitions were recorded with the individual in the standing position. Kinematic motion analysis involved selecting scapular and humeral data at rest, 30°, 60°, 90° and 120° of arm elevation. The results were analyzed using the SPSS statistical package. For scapular internal rotation, upward rotation, posterior tilt and humeral superior and anterior translations a 2-way ANOVA was used, in separate, to test interactions of angle x group or main effect of group. Bonferroni post-hoc test was used when necessary. The level of significance was considered 5%.

RESULTS

No angle x group interactions were found for scapular kinematic variables or humeral superior translation. The main effect of group was only significant for scapular tilt (p=0.04; F=2.78). Individuals with shoulder pain and without PCT presented decreased anterior tilt when compared to individuals without pain and with PCT (Figure 1). There was an angle X group interaction for anterior translation (p=0.014; F=2.84). Individuals with shoulder pain and without PCT showed decreased anterior translation when compared to the other groups (Figure 2).
DISCUSSION AND CONCLUSIONS
The results of the present study were partially supported by the literature that has shown that PCT may contribute to increased scapular internal, downward and anterior rotation, and anterior and superior humeral translations.\(^7\)\(^\text{-}^{10}\) It was hypothesized that the groups with shoulder pain would present increased scapular internal rotation, and decreased scapular upward rotation and posterior tilt when compared to the asymptomatic group independently of the presence of PCT. It was expected also that the alterations would be more prominent in the group with pain and PCT. However, only the group with tightness without pain showed increased anterior tilt as previously described in asymptomatic throwers.\(^7\)\(^,\)\(^9\)

The fact that past studies\(^7\)\(^\text{-}^{10}\) have only evaluated asymptomatic subjects with PCT may explain the divergent results between our study and the previous ones concerning the humeral motion. The present study showed decreased anterior translation of the humerus in individuals with pain and without PCT. Studies assessing scapular and humeral motions in individuals with PCT and concomitant shoulder pain were not found in the literature search. Considering this, direct comparison of the current results to the previous investigations cannot be made. Further investigations are necessary in subjects with PCT with and without shoulder pain to clarify the real changes caused by PCT.

REFERENCES

ACKNOWLEDGEMENTS
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Scapular kinematics in able-bodied subjects during manual wheelchair propulsion

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Introduction:
In comparison to the general population, the prevalence of shoulder pain in persons with a spinal cord injury SCI is high (30-70%) (Curtis et al., 1995; Luime et al., 2004). Shoulder pain and disorders affect persons with spinal cord injury (SCI) since they are dependent on the upper extremity for mobility, participation and activities of daily living. In earlier research, shoulder pain has been associated with altered scapular kinematics during static poses and arm elevation (Timmons et al., 2012). A narrowing of the subacromial space and subsequent impingement of the structures in the subacromial space has been associated with reduced lateral rotation and posterior tilt of the scapula.

To prevent shoulder complaints in persons with SCI, early diagnosis and treatment of disorders, such as dyskinesia, are required. Following SCI, there is a large amount of variety in the upper body movement pattern due to decreased flexibility of the spinal column and paralysis of trunk and arm muscles (Desroches et al., 2013; Gauthier et al., 2013). It is reasonable to assume that differences exist between persons with SCI and able-bodied persons.

In order to diagnose dyskinesia in persons with SCI, the healthy movement pattern of the scapula needs to be well understood.

The primary aim of this study is to describe scapular kinematics in able-bodied persons in a standardized static pose, during arm elevation and manual wheelchair propulsion.

Methods:
Sixteen able-bodied novices (10 males, 6 females) with a mean age of 25.4±8.3 years, mean mass of 71.1±13.8 kg and mean height of 1.79±0.09 m, voluntary participated in this study. Exclusion criteria were any prior experience with manual wheelchair propulsion, intensive regular upper-body exercise or medical contra-indications.

All subjects performed a standardized pose in the anatomical position, arm elevation in the frontal plane and manual wheelchair propulsion on a motorized treadmill, during three four-minute steady-state low-intensity (0.25W/kg body weight) practice blocks. Kinematic data of the upper body were collected with an optoelectric camera system combined with skin-mounted cluster markers and kinetic data were collected using an instrumented wheelchair wheel during all practice blocks. In accordance with the ISB guidelines, bony landmarks were probed and accordingly reconstructed. Subsequently, Euler angles of the joints of the upper body were reconstructed.

Results:
In the anatomical position, 34.7° (SD 5.3°) of scapular protraction, 6.8° (SD 9.2°) of lateral rotation and 9.3° (SD 5.8°) of anterior tilt of the scapula were observed at 18.7° (SD 10.0°) of arm elevation. At 80° of arm elevation in the frontal plane, scapular protraction and posterior tilt were lower at respectively 34.3° (SD 7.4°) and 3.2° (SD 7.4°), medial rotation increased to 28.8° (SD 5.3°).

Scapular kinematics during the push phase of manual wheelchair propulsion for respectively protraction, lateral rotation and posterior tilt were found to be on average 32.7° (SD 5.3°) of scapular protraction, 6.8° (SD 9.2°) of lateral rotation and 9.3° (SD 5.8°) of anterior tilt of the scapula were observed at 18.7° (SD 10.0°) of arm elevation. At 80° of arm elevation in the frontal plane, scapular protraction and posterior tilt were lower at respectively 34.3° (SD 7.4°) and 3.2° (SD 7.4°), medial rotation increased to 28.8° (SD 5.3°).

Scapular kinematics during the push phase of manual wheelchair propulsion for respectively protraction, lateral rotation and posterior tilt were found to be on average 32.7° (SD 5.3°), 7.1° (SD 5.2°) and 9.8° (SD 5.3°). Their respective ranges were of 15.6° (SD 5.2°), 8.8° (SD 7.5°) and 10.1° (SD 4.6°). An overview of the scapular kinematics during an average, standardized push phase is shown in Figure 1.

Discussion:
The bone-pin study of Ludewig and colleagues shows comparable results in the static posture of persons with low-level paraplegia (Ludewig et al., 2009). Their findings were 41.1° of protraction,
5.4° of lateral rotation and 13.5° of anterior tilt. Scapular kinematics during arm elevation are in agreement with earlier research in able-bodied persons (Barnett et al., 1998). The movement pattern of the scapula shows similarity to the pattern found in persons with SCI, however the protraction and posterior tilt are systematically 10° lower.

In summary, the observed movement pattern of the scapula in able-bodied individuals shows good comparability to prior research and provides a base for investigating dyskinesia in persons with SCI.

References:

**Figure 1.** Mean scapulothoracic joint kinematics during a standardized push cycle. The bold line indicates the mean of all per-subject means over 4 complete propulsion cycles. The grey area encloses 1 standard deviation from the mean. On average, a push-phase was 38% of the complete propulsion cycle.
3D RECONSTRUCTION OF SCAPULA FROM BIPLANE X-RAY IMAGES

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INTRODUCTION
Three-dimensional (3D) reconstruction from biplane X-ray images may reduce radiation compared to CT scan and allow acquisitions in functional positions. To reconstruct the scapula, a method using parametric model with statistical inference information has been proposed [1]. This method provided an initial solution with further manual adjustments, which is adapted to different object orientations and calculation of clinical relevant parameters. However, the method is time-consuming. Therefore, the aim of this research is to propose a fast reconstruction method of the scapula, based on an improvement of the initial model and automatic adjustments using signal processing.

METHODS
Generation of scapula model: reduced presentation and statistical analysis
Forty scapulae were segmented in CT images and transformed into the local anatomy coordinate system with global scaling based on the distance between the inferior angle and trigonum scapulae points as demonstrate in Figure 1. Landmarks were either selected manually or calculated automatically from manually selected regions by an expert surgeon. A contour model as a reduced representation of the scapula was calculated with three contours and seven landmarks. The three contours were the lateral border, medial border and inferior line of the supraspinatus fossa, that were represented as 140 equally positioned key points generated by cubic cardinal spline fitting from original contour extraction. The three contours were selected because they represent the scapula body. Seven point landmarks, which were the posterior, anterior and center of spine surface, anterior of acromion root, superior border of glenoid, center of acromion root and coracoid tip, were selected to represent the other main regions of the scapula. To obtain the contour model, a principal component analysis was applied on the 3D coordinates of the entire 147 key points. The average shape as well as shape constraints of the first six principal components were used in the next step. Based on the visibility of the key points on subsequent biplane X-ray images, 140 key points were used for automatic adjustment and 3 key points were identified on the images.

Figure 1: Generation of scapula model: one scapula surface with extracted landmarks (red points) and contours (black point). Black dot line: line linked GC and TS. TS, CG and AI points were used to generate the local coordinate system. Blue cycles were the key points of the contour model used for automatic adjustment. The three arrow shows the local glenoid coordinate system.

Biplane X-ray image acquisition
Biplane X-ray images were obtained for six subjects, using the low-dose EOS® system (EOS Imaging, Paris, France). Each acquisition produced simultaneously two calibrated X-ray images in two orthogonal planes with subjects kept at standing position with an axial rotation of about 45°.

Automatic adjustment
The 3D orientation of lateral border can be calculated since it was a stereo-corresponding contour (SCC) that can be identified on both X-ray images. On the contrary, medial border and inferior line of the supraspinatus fossa were non-stereo-corresponding contours (NSCC), which can only be identified on one image. Therefore, the NSCC plans were generated as sets of back projection X-ray lines (Figure 2). Hence, the contour model was first size registered and pre-orientated based on the length and 3D orientation of the lateral border by applying iterative closest point (ICP) method. The pre-dimensioned and pre-oriented contour model was marked as the initial model. In the second step, the initial model rotated rigidly through the axis of lateral border to approximate the NSCC planes until the total distances of the key points were minimized.
third step, landmarks on the medial border and inferior line of the supraspinatus fossa were deformed elastically to approach the NSCC plans by respecting the shape constraints. By adding two stereo-corresponding points (SCP), which were the center of acromion root and the center of coracoid tip and the projection point of inferior border of glenoid, identified manually on the images, the personalized contour model was obtained.

Finally, to get the entire personalized scapular shape, a mesh surface model was deformed using the personalized contour model and control points. It included two steps, a global deformation using moving least squares and a local deformation using dual kriging interpolation. The scapula plane was then calculated as the least square plane of the three borders.

Figure 2: The EOS system with two X-ray sources and two detectors, illustrated with the scapula model and back projection planes of the two non-stereo-corresponding Contours.

Data analysis
To analyse the reconstruction result, the personalized scapula model was compared to the average manual reconstructions performed by two different experts for 12 subjects. The orientation of the scapula body and the root mean squares (RMS) and maxima of the point surface differences of glenoid surface were compared.

RESULTS
Time required for image segmentation of one scapula and automatic adjustment was less than three minutes compared to 20 minutes for manual reconstruction. The differences of scapula plane orientation between the two approaches were less than 2.5° (Table I) except one subject that reached to 3.4°. For the point surface differences of glenoid surface, inter-operator differences among the two reconstructions using the manual approach were less than 2.1mm (RMS) and 4.5mm (maxima), except for one subject. And the differences between the two approaches were less than 2.1mm (RMS) and 4.3mm (maxima) for 11 subjects among 12.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Scapula orientation (°)</th>
<th>Point-surface differences (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Inter-operator variations between two approaches</td>
<td>Inter-operator variations between two approaches</td>
</tr>
<tr>
<td></td>
<td>RMS</td>
<td>MAX</td>
</tr>
<tr>
<td>1</td>
<td>2.8</td>
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</tr>
<tr>
<td>2</td>
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<td>12</td>
<td>2.5</td>
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</tbody>
</table>

Table 1: Inter-operator variations of the manual method and comparison between two approaches for the scapula orientation and point surface differences of the glenoid. RMS: Root means squares; MAX: Maxima.

DISCUSSION AND CONCLUSIONS
We proposed an improved method of 3D scapula reconstruction based on reduced representation of the scapula and automatic adjustments using signal processing. The technique may generate reliable scapular orientation and the time required for one reconstruction is short. For glenoid surface generation, 11 subjects among 12 presented differences between the present method and the previous approach smaller or similar to the inter-operator variation. One subject had high inter-operator variation because of the low image quality. However, this research is limited by the lack of CT scan validation. Thus, the next step is to validate the method on digitally reconstructed radiograph with CT image segmentation.

REFERENCES

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A new glenohumeral joint model for the study of shoulder stability

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INTRODUCTION
The shoulder has a large range of motion, which is achieved at the cost of stability. The rotator cuff is a group of muscles whose primary role is to stabilize the glenohumeral (GH) joint, while other shoulder muscles such as the deltoid are the primary joint movers. Musculoskeletal modeling can be used to estimate muscle forces required for observed motion, but in the case of the shoulder it is important to take into account both the mobility and stability of the joint, otherwise the estimated forces could result in shoulder dislocation.

Our overall aim is to build a shoulder assessment tool that will assist clinicians in the design of optimal shoulder interventions. This tool will consist of a combination of motion analysis and musculoskeletal modelling. For the model to be suitable for use in clinical practice, it needs to be easy to use, and accurately represent how individual muscles drive the motion of the shoulder.

We are using OpenSim to model and simulate the mechanics of the shoulder. OpenSim is a user-friendly, open source musculoskeletal modeling and simulation package [1]. The OpenSim tool that runs inverse-dynamic simulations and calculates individual muscle forces is Static Optimization. It solves the equations of motion and distributes the forces to the muscles by minimizing muscle activations, but does not take into account GH stability.

Therefore, the objective of this study was to create a new OpenSim Static Optimization tool that includes the competing goals of minimum activation and glenohumeral stability. We aim to evaluate the new tool using electromyography data from rotator cuff muscles recorded from healthy volunteers.

METHODS
We have previously built a detailed shoulder model in OpenSim that includes all joints and muscles based on cadaver measurements [2]. We have added to this a new model of the scapulothoracic joint, described in [3]. We describe GH stability using the joint reaction force vector: the joint is assumed to be stable when the direction of this vector lies inside the rim of the glenoid, modelled as an ellipse, and instability is assumed to arise when the vector falls outside the rim (figure 1). The size of the ellipse describing the glenoid is taken from cadaver studies [4]. The GH stability value is defined as:

\[
GH_{\text{stability}} = \left(\frac{\theta}{\theta_a}\right)^2 + \left(\frac{\phi}{\phi_a}\right)^2
\]

where \(\theta\) and \(\phi\) are the angles of the vector away from the normal to the glenoid along the major and minor axes of the ellipse representing the glenoid fossa, and \(\theta_a\) and \(\phi_a\) are the angles of that vector at the rim of the glenoid. GH\text{stability} is 0 at the centre of the glenoid fossa, and approaches 1 as the vector approaches the edge of the fossa.

Figure 1: The orientation of the glenohumeral joint reaction force with respect to the ellipse representing the glenoid fossa of the scapula

We include GH stability as a second term in the Static Optimization cost function:

\[
J = \sum_{i=1}^{n_x} x_i^2 + \omega GH_{\text{stability}}^2
\]

where \(x_i\) are the muscle activations, and \(\omega\) is a weighting factor set to 1.
Five healthy volunteers (aged 25-45) underwent motion analysis while performing a set of tasks including standard arm elevations, simulated activities of daily living and force tasks designed to challenge the stability of the GH joint. 3D kinematics of the arm (Vicon Ltd), muscle activity (Delsys Trigno), and hand forces (AMTI) were recorded simultaneously. Fine-wire recordings of EMG were taken from infraspinatus, supraspinatus and teres minor.

These data were used as inputs to the inverse dynamic simulations carried out with the OpenSim model. We calculated the linear envelope of the normalized EMG for the three rotator cuff muscles, and compared it to the muscle activity predicted by the modified Static Optimization and standard Static Optimization using the root mean square error.

RESULTS
We show preliminary results from one subject, for a 3.5s isometric force trial during which the shoulder was at 90 degrees of flexion. The subject pushed downwards maximally against a force sensor between the times of ~1.2 and 3.2 seconds.

Figure 2 shows a sudden increase in GH instability at the time of force application when the standard Static Optimization is performed. Including the GH instability as a factor in the cost function keeps joint much more stable.

Figure 3 shows the agreement between measured and modelled muscle activities for the sample movement. The RMS error is lower for each muscle where GH stability is included in the optimisation.

DISCUSSION AND CONCLUSIONS
Our preliminary results show that the modified Static Optimization tool leads to prediction of much greater stability in the GH joint in a stability-challenging task. The standard Static Optimisation allows the joint to become unstable, which is not observed in the participants performing the movement. Differences between modelled and measured muscle activations may also be reduced with the new tool, improving model validity, although at this stage only limited data from a single subject is shown. Analysis of the complete dataset will be required to confirm these initial findings. More complete validation is planned in future work that will compare estimated GH contact forces with those from an instrumented shoulder prosthesis.

REFERENCES

ACKNOWLEDGEMENTS
The authors would like to acknowledge funding from the National Institutes of Health (R24 HD065690).
Relationship between the Axial Rotational Range and Bone Geometry of the Glenohumeral Joint

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INTRODUCTION
The maximum internal and external rotation of the Glenohumeral joint (GHJ) shows large variation between individuals and is dependent on the humeral elevation angle and elevation plane of the humerus [1]. This could be due to differences in the bony and soft-tissue restraints of the joint between individuals. For example, it has been previously reported that baseball athletes who have a greater angle of retroversion of the humerus can achieve a greater external rotation and reduced internal rotation of the GHJ at higher humeral elevation angles [2]. Also, it has been shown that injury and pathology of the GHJ joint compromises the range of motion of the shoulder [3]. Scans of the shoulder have been used to create bone models to predict the maximum range of motion of the joint. This detects the rotation angles for which collision of the humerus and scapula bones occur [4], thus this can optimise surgical outcomes in patients with shoulder pathology. This study aims to investigate the relationship between shoulder bone geometry and axial range of motion of the GHJ. The study first provides a comprehensive description of the range of axial rotation. Next, bone geometrical parameters of the GHJ acquired from MRI scans are correlated to the range of axial rotation to improve understanding of the complex movement of the GHJ. The study’s long term goal is to assist in the development of shoulder implants, pre-operative planning for shoulder surgery and classification of shoulder disorders.

METHODS
Measurement of the GHJ range of axial rotation
Kinematic data of the GHJ was collected using optical motion tracking from 10 healthy subjects, who had no history of shoulder pathology. The experimental set-up used to quantify the GHJ active and passive axial rotational range at various humeral positions has been previously developed and assessed [1] (Figure 1). The maximum internal and external rotation was measured at 60º, 90º and 120 º humeral elevation in the Coronal, Scapular and Sagittal planes during active motion and passive rotation in response to a measured torque.

A 6 degrees of freedom marker set tracked motion of the scapula and humerus, where bony landmarks and segment coordinate systems were defined according to ISB recommendations [5]. Rotations of the GHJ were quantified using Euler sequence YX’Y’’ [5].

![Figure 1](image1.png)

**Figure 1**: The experimental set-up used to quantify the maximum internal-external rotation of the GHJ at multiple humeral positions during active and passive motion [1].

Bone geometrical parameters of the GHJ
Bone geometrical parameters of the humerus and scapula were measured from MRI scans of the 10 subjects. A scan of the scapula and proximal humerus was acquired with a resolution of 1mm and a scan of the whole humerus was acquired at 5mm resolution. Scans were segmented in ScanIP (Version 4, Simpleware, UK).

Geometrical parameters of the glenoid were measured in the plane of the scapula in the slice with the greatest glenoid height (Figure 2); whilst the shape of the humeral head (diameter, inclination, retroversion) was measured in the slice with the greatest humeral head diameter. Geometrical parameters of the articular cartilage (height, diameter and curvature) were acquired in the plane of the scapula in the slice which showed greatest definition of the articular cartilage. The acromion shape (length and subacromial depth) was measured in the slice in the Sagittal plane with greatest acromion length.

![Figure 2](image2.png)

**Figure 2**: 2D geometrical parameters of the glenoid (depth (i, ii, iii), height (iv) and curvature (v)) measured in a slice in the anatomical plane of the scapula.
Relationship between GHJ axial range of motion and bone geometry

A matrix of Pearson product moment correlation coefficients determined which geometrical parameters showed strongest correlation with the maximum internal-external rotation. Weighted linear regression expressions combined up to three geometrical parameters at each humeral position to predict maximum internal and external rotation of the GHJ. The expressions were validated using a leave-one-out experiment. Segmented images were used to create subject-specific bone models and regions of bony collision were identified in SolidEdge (Version 7, Siemens, UK). The humerus was rotated to maximum internal and external rotation in the model at each humeral position using angles quantified from the subject’s kinematic data of the GHJ. The humerus was allowed to translate by up to 3mm from the glenoid in three directions when simulating passive rotation.

RESULTS

During active rotation, the shape of the glenoid and articular cartilage affected maximum internal-external rotation. In the Scapular plane, parameters which showed strongest correlation with the maximum internal rotation were the glenoid curvature and depth. These parameters were more strongly correlated to maximum internal rotation at 60º elevation, where a greater active internal rotation was achieved with greater glenoid curvature ($r = 0.76$, $p < 0.001$) and a greater inferior glenoid depth ($r = 0.76$, $p < 0.001$). This was supported in the bone model, as collision between the lesser tuberosity and inferior edge of the glenoid limited maximum internal rotation at this humeral position. A greater height of the articular cartilage allowed a greater external active rotation to be achieved at 60º elevation in the Scapular plane. The region of articular cartilage available to contact with the glenoid was shown to limit maximum external rotation, illustrated by the blue line in Figure 3a. A greater articular cartilage height increased the articular cartilage coverage and enabled a greater external rotation to be achieved.

During passive rotation, the shape of the acromion was more likely to limit maximum internal-external rotation. In the Scapular plane, the glenoid depth and subacromial depth showed strongest correlation with the axial rotational range. These parameters were more strongly correlated at 60º elevation ($r = 0.87$, $p < 0.01$) and 90º elevation ($r = 0.66$, $p < 0.05$) respectively. Upon collision between the humeral head and glenoid, anterior-posterior translation of the humeral head allowed further axial rotation to be achieved before collision between the humeral head and posterior edge of the acromion (Figure 3b).

At 90º elevation in the Scapular plane, passive internal rotation was predicted using Eq. 1. The leave-one-out experiment showed no significant difference between quantified maximum internal and external rotations and values predicted from linear regression expressions at each humeral position.

$$\text{Passive internal rotation at 90º elevation in the Scapular plane} = 1.3 \times \text{Subacromial depth} + 2.0 \times \text{Glenoid curvature} - 224.9 \quad \text{Eq.1}$$

DISCUSSION AND CONCLUSIONS

Strong correlations between geometrical parameters and maximum internal-external rotation were supported by observations from collision detection. This improves understanding of bony constraints of the GHJ and shows variation in axial rotational range is affected by variation in bone shape between individuals. The results suggest linear regression models can be used to predict axial rotational range of the GHJ using geometrical parameters acquired from MRI scans, thus assisting in defining rehabilitation targets.

REFERENCES


ACKNOWLEDGEMENTS

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Analysis of Soft Tissue Artefact using the Optimal Common Shape Technique during Glenohumeral Internal Rotation

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INTRODUCTION
Measurement of glenohumeral kinematics with retro-reflective markers is problematic due to soft tissue artefact (STA). STA affects axial rotation of the humerus (internal and external rotation) in particular and can result in significantly altered estimates of rotation (1). A cluster of markers is often attached to the upper arm to minimize the impact of STA associated with tracking of the medial and lateral humeral epicondyles. However, STA is likely to vary across the segment making choice of marker location critical for accurate determination of humeral kinematics.

The Optimal Common Shape Technique (OCST) is an optimisation technique that accounts for local skin deformation through the use of Generalised Procrustes analysis (GPA) and provides a rigid marker cluster configuration reducing STA (2) and improving the precision of functionally identified joint centres (3,4). Moreover, comparison of a rigid reference marker cluster configuration determined by the OCST to the measured positions of a grid of markers on the upper arm through Ordinary Procrustes analysis (OPA) can provide an estimate of the amount of STA present at various sites on the bone. Such STA estimates can help to identify optimal marker locations to improve the estimation of humeral kinematics. The aim of the study was to assess the utilisation of the OCST in the measurement of humeral kinematics through the quantification of the STA and its effect on estimating humeral axial rotation.

METHODS
Thirteen wheelchair users, three of which were recreationally active wheelchair rugby players and 10 who were wheelchair tennis players, were recruited. Kinematics of the thorax, scapula, humerus and ulna were captured using a Vicon Motion Capture system (Oxford, UK). Retro-reflective markers were attached to the thorax and forearm in accordance with ISB recommendations (5) and an acromion marker cluster was attached to track the scapula (6). A cluster of markers was attached to the upper arm with each marker attached directly to the skin. Four markers were attached bilaterally on the anterior, lateral and posterior sides of the upper arm spaced equidistant starting distal to the deltoid muscle and ending approximately 3cm proximal to the elbow joint line. Markers were labelled A1 to A4, L1 to L4, and P1 to P4 from proximal to distal on the anterior, lateral and posterior aspect respectively (Figure 1).

Figure 1: Upper arm marker placement. Marker labels for the most proximal and distal markers on the anterior (A1 and A4), lateral (L1 & L4) and posterior (P1 & P4) sides.

Marker data was firstly recorded during a static pose defined as 0° of humeral elevation, elbow fully extended and wrist supinated. Participants then performed three repetitions of maximal active humeral internal rotation. Starting with their arm abducted to 90° and elbow flexed to 90° participants actively internally rotated their arm until they reached their maximum whilst maintaining 90° arm abduction.

The optimal common shape of the upper arm marker cluster was determined through the use of the GPA during the static pose. The OPA was then used to fit the optimal common shape to the marker data of the internal rotation movements. STA of each upper arm marker was then determined as the Euclidian distance between the raw and OPA fitted marker positions and the 3D displacement vectors of the raw markers with respect to OPA fitted markers.

In order to determine the effect of STA on kinematics a comparison was made between Option 1 and Option 2 of the ISB recommendations in defining the local coordinate system of the humerus. Option 1 utilizes the plane formed by the glenohumeral joint centre and the medial (EM) and lateral (EL) humeral epicondyles. Axial rotation of the humerus using Option 1 is dependent on the rotations of the EM and EL. The location of the EM and EL landmarks are
determined based on their relationship with the upper arm marker cluster and are, therefore, susceptible to STA. Option 2 utilizes the plane formed by the glenohumeral joint centre, elbow joint centre and the ulna styloid. Option 2 is less susceptible to STA when determining axial rotation as it does not rely on the EM and EL landmarks in providing the axial rotation movement. However, the elbow joint centre is identified with respect the EM and EL landmarks and may still be subject to STA. The glenohumeral joint centre was determined between the acromion marker cluster and the OPA fitted upper arm marker configuration using the SCoRE method (3). A YX’Y” non-cardan rotation sequence was used to estimate humeral rotation with respect to the thorax (7). The range of motion was calculated for Option 1 and Option 2 for the OPA fitted marker configuration. Data from the left and right arms were combined resulting in a sample size of 26 arms.

Following analysis of the STA a revised marker configuration was defined consisting of the A4, M4 and P4 markers. Post-processing of data including OCST optimisation, determination of the glenohumeral joint centre and estimation of axial rotation was repeated with the revised marker configuration.

A repeated measures ANOVA with main effects of ISB option (2 levels) and marker configuration (2 levels) was used to test for statistical significance.

### RESULTS

The average Euclidian distance between raw and OPA fitted markers was greatest for the proximal and distal markers with an average distance of 8mm, 18mm, 22m for the A1, L1 and P1 markers, and 18mm, 10mm and 14mm for the A4, L4 and P4 markers. The average distance for the A2, A3, L2, L3, P2 and P3 markers was 9mm, 9mm, 9mm, 4mm, 10mm and 7mm respectively.

The displacement of the raw markers with respect to the OPA fitted markers showed the A1 marker moved 4mm anteriorly and 6mm laterally and the P1 marker moved anteriorly and medially 15mm and 18mm respectively. The A4 marker moved 8mm anteriorly and 17mm medially. The P4 marker moved 14mm posteriorly and 12mm laterally. Interpretation of these movements suggests the proximal markers are underestimating axial rotation, therefore, the revised marker configuration consisting of only the distal markers was assessed.

There was a significant (P<0.001) underestimation of internal rotation between Option 1 and Option 2 by 45° when using all markers on the upper arm (Table 1).

The revised marker configuration, utilising only the distal markers, estimated a significantly (P<0.001) larger range of motion compared to using all markers when the humerus was defined using Option 1 (Table 1). The revised marker set significantly underestimated range of motion when compared to Option 2 by 23°.

### DISCUSSION AND CONCLUSIONS

The OCST based technique highlighted large STAs that varied in magnitude across the length of the upper arm. Defining the humerus using Option 2 would be the preferred option when estimating humeral kinematics, however, this method is dependent on elbow flexion being present and is affected by hand pronation/supination. In conditions where only the upper arm markers need to be used to estimate humeral internal rotation the underestimation in axial rotation can be reduced through assessment of the STA and subsequent selection of markers on the distal upper arm. This study represents the first use of the OCST in optimising marker configurations for the upper limb and provides an easily applicable method for constraining inter-marker distance and can be used to determine optimal marker location to improve estimations of humeral kinematics.

### REFERENCES


### ACKNOWLEDGEMENTS

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Anatomical reconstruction of glenoid bone defects using a statistical shape model

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INTRODUCTION

A common method to treat patients with an arthritic glenoid is to reconstruct the original glenoid surface in order to restore the function of the shoulder joint. Glenoid medialisation and glenoid orientation are generally known to affect stability and range of motion of the shoulder joint. Quantification of the glenoid bone defect and assessment of its original orientation are therefore important factors in the surgical planning of shoulder joint replacement. Pre-operative planning software supports the surgeon in assessing the original glenoid shape of the patient. The planning software should therefore include a virtual anatomic reconstruction of the patient’s bone defect. A common method of bone reconstruction uses the contralateral bone to assess the original shape of the defect bone [1, 2]. Glenoid arthritis however, is often reported at both sides which makes this method inapplicable. Moreover, a scan of the contralateral side is not always present. A reconstruction method based on a statistical shape model (SSM) was already described in literature for the pelvis [3]. The goal of this study is to verify whether an SSM of the scapula is accurate enough to reconstruct the glenoid bone defect.

First, we describe the construction of the SSM. We then apply the SSM-based reconstruction technique and assess the performance of the reconstruction method for use in preoperative planning of shoulder surgery.

METHODS

SSM creation

A database of 28 healthy scapula was used as a training set to construct the SSM. The scapula scans were segmented in the image processing software Mimics (Materialise, Leuven, Belgium) and converted to a 3D triangular mesh. The template-based registration method described in Danckaers et al 2014 [4] was applied to obtain a corresponding mesh in all samples of the database. A Procrustes analysis was performed to exclude the translational and rotational variations between the samples. Finally an SSM was created using Principal Component Analysis (PCA) on the meshes in the dataset.

SSM-based reconstruction

The workflow used for the reconstruction of a deformed scapula is illustrated in Figure 1. In a first step, the defected parts are manually removed using the modelling software 3-matic (Materialise, Leuven, Belgium). In a next step, the SSM is fitted to the remaining healthy structures of the scapula using the fitting algorithm described in Vanden Berghe et al 2016 [3]. The modes of variation are optimized one by one to minimize the distance between the SSM and the healthy parts of the original scapula.

Reconstruction performance

Artificial defect regions were cut out in each healthy scapula of the dataset. A mild, a severe and an extreme defect region were indicated to account for multiple defect types (Figure 2). The performance of the SSM-based reconstruction was assessed by a leave-one-out approach: For each sample, an SSM was created with all other samples in the dataset. This new SSM was then used to reconstruct the artificial defect region of the left-out sample. The fit error was defined as the mean Euclidian distance between the best SSM-fit and the healthy part of the original sample (blue in Figure 2). The reconstruction error was defined as the mean Euclidian distance between the best SSM-fit and the artificial cut-out defect (red in Figure 2). To assess the applicability for pre-operative planning of shoulder surgery, two clinically relevant parameters were defined: the glenoid plane orientation and the glenoid centre. The glenoid plane was determined as the best-fit plane through the points on the glenoid rim (Figure 3). The glenoid centre was calculated as the centre of the circle that fits the inferior glenoid rim [5]. To further evaluate the reconstruction performance of the SSM, the error
on the glenoid plane orientation and the glenoid centre was measured between the best SSM-fit and the original sample.

FIGURE 2: Three different defect regions (red) are cut out in each scapula, representing a mild defect, a severe defect (including coracoid) and an extreme defect (including acromion and coracoid).

RESULTS
Figure 3 shows the mean shape of the scapula SSM. The average fit error, the average reconstruction error and the average error on the glenoid orientation and centre are given in Table 1.

Table 1: Mean and standard deviations of the errors

<table>
<thead>
<tr>
<th></th>
<th>Fit error (mm)</th>
<th>Rec. error (mm)</th>
<th>Glenoid orient. Error (°)</th>
<th>Glenoid centre error (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mild</td>
<td>0.89 (0.15)</td>
<td>1.34 (0.41)</td>
<td>4.9 (3.0)</td>
<td>2.73 (1.17)</td>
</tr>
<tr>
<td>Severe</td>
<td>0.86 (0.14)</td>
<td>1.72 (0.48)</td>
<td>5.0 (2.9)</td>
<td>2.74 (1.15)</td>
</tr>
<tr>
<td>Extreme</td>
<td>0.76 (0.14)</td>
<td>2.02 (0.61)</td>
<td>6.0 (3.7)</td>
<td>3.78 (1.33)</td>
</tr>
</tbody>
</table>

DISCUSSION AND CONCLUSIONS
The results in this study show that an SSM of the scapula is able to reconstruct a glenoid defect with a surface accuracy of 2.0 mm or less. The glenoid centre and the glenoid orientation can be predicted with a minimal accuracy of 3.8 mm and 6° respectively. In small glenoid bone defects, these predictions will be more accurate than in large glenoid bone defects. The fit error on the healthy parts is in between 0.7 and 0.9 mm. The decreasing fit error for increasing defect sizes can be explained by the size of the healthy region. When the remaining healthy region becomes small, the fit algorithm will find a more accurate fit to that region.

The reported reconstruction errors are comparable to the errors that occur in hip reconstruction: Using a similar technique, Vanden Berghe et al 2016 [3] reported a reconstruction error of 1.5 mm and an acetabular centre error of 3.2 mm. However, the acetabular orientation error in Vanden Berghe et al 2016 [3] was only 3.0° compared to an observed glenoid orientation error of 6°. This may be attributed to the fact that a glenoid is less constrained by the rest of the bone compared to an acetabulum. The shape variations in the glenoid may therefore be less predictable by the remaining scapula. Compared to the contralateral method however, the SSM-based reconstruction method allows to reconstruct a glenoid defect without the need of a healthy contralateral bone.

Future work on this topic will focus on exploring improvements that further increase the accuracy of the SSM-based reconstruction method for the scapula. In that context, the influence of a larger scapula dataset will be investigated. Furthermore, we will allocate an SSM-based scapular coordinate system to specifically assess the inclination and version of the original glenoid surface. Despite these improvements, the current study already shows that an SSM-based reconstruction method is a promising tool for virtual glenoid reconstructions.

REFERENCES

ACKNOWLEDGEMENTS
This work has financial support by the Baekeland scheme of the agency for Innovation by Science and Technology (IWT).
INTRODUCTION
Shoulder mobility is an important clinical sign in the assessment of the shoulder joint [9]. However, there is no gold standard measuring shoulder mobility in clinical practice. While some health care personnel use functional scales like the Constant-Murley [4], others use a portable goniometer, and scientists use advanced motion capture systems (Mocap) to measure the range of motion (ROM) [7]. Unfortunately, the Constant-Murley Scale tells nothing about the ROM as it analyses functional mobility [4]. Although low-cost and quick to perform, a goniometer can only be used in the two dimensional area, needs trained personnel, and is related to a measurement error of 8° [2, 3]. Basically, visual estimation of the ROM results in the same error, but depends on the expertise of the examiner and the pain intensity of the patient [9]. In contrast, advanced systems like X-ray machines, cinematography or Mocaps allow to analyse shoulder joint motion in the three dimensional area, resulting in a significantly smaller measurement error [6]. However, they are time consuming, need highly trained personnel and an advanced laboratory equipment making them unsuitable for clinical use. A shoulder patient passes therefore several mobility measurement types in his course of therapy depending on the preference of his health care personnel.

It would therefore be very attractive to have a low-cost, easy-to-use, three dimensional measuring device to asses shoulder mobility in applied settings. In addition, it would be very smart if anyone, even without preliminary knowledge in the field of shoulder measurements, could use this device intuitively. For this reason, smartphone applications (Apps) were developed. Since existing Apps need trained personnel [7, 10], have no exercise instructions [8], or show incorrect measurements [8, 11], the ZHAW developed a new therapy application (therapp®, available in German speaking AppStores®). The program uses the factory implemented gyroscope of the iPhone® to analyse three dimensional motions and shall allow the user to measure the own shoulder ROM without preliminary knowhow (Figure 1).

Figure 1: Subject showing the Velcro to place the smartphone at the upper arm (left) and while performing a shoulder flexion under guidance of the App (right).

METHODS
Study population
Classical scenario-based usability studies include at least five but ideally ten subjects [1, 5]. As no preliminary data for validity were available, sample size determination for usability was used. We therefore included 12 subjects, half of them without preliminary knowledge (lays) and half of them physiotherapists (pros). Only healthy subjects (aged between 25 to 65 years) with an own smartphone were recruited (mean age 35 years).

Tasks
Every subject performed three tasks in randomized order under guidance of the App, each task consisting of 3 repetitions: shoulder abduction (ABD), shoulder flexion (FLEX), and shoulder external rotation (ER) with a 90° flexed elbow. Only the first valid repetition of each task was evaluated.

Data gathering
A formative usability evaluation with an inductive test design (laboratory-based method of thinking aloud) was performed [5]. Test sessions were videotaped and smartphone displays simultaneously transferred to the examiner computer. Participants were not instructed on how to use the App. After completion of the tasks, participants filled out a questionnaire. Concurrent validity was tested using the factory implemented upper body model of the Mocap.
Sensors were placed according to manufacturer’s recommendation using double-sided adhesive tape.

**Data analysis**

For usability evaluation, a qualitative content analysis approach was used (inductive category formation with focus on content structuring). Video tapes were transcribed in their full length. Barriers were further identified in an iterative process, categorized and weighted by four experts (2 lays and 2 pros) regarding severity. Two examiners further evaluated importance of troubleshooting of each barrier and discussed every item until final consensus was found. At last, combination of weighting and troubleshooting was used for revision recommendations.

For validity evaluation, ROM of the shoulder was analysed in the primary motion plane. The App transferred quaternions into Euler angles (X, Y', Z''). For the Mocap, the shoulder angle was determined using the implemented biomechanical model. Directional vectors were defined based on body landmarks, and analysed regarding ROM. Absolute bias between the two methods was analysed.

**RESULTS**

All subjects were able to perform the three tasks, but one subject was not able to find the results (ROM) in the App. Furthermore, 78 barriers were identified, whereas 8 were categorized as catastrophe, 10 as major and 17 as minor issues, 35 as only cosmetically and further 8 as no usability problem. In combination with troubleshooting, for 29 barriers a revision is recommended.

Table 1: Range of motion (mean±sd) for the two systems and the bias (Mocap minus App) between systems.

<table>
<thead>
<tr>
<th>task</th>
<th>App</th>
<th>Mocap</th>
<th>bias</th>
</tr>
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<tbody>
<tr>
<td>ABD</td>
<td>100±51°</td>
<td>137±30°</td>
<td>-37±41°</td>
</tr>
<tr>
<td>FLEX</td>
<td>104±57°</td>
<td>157±16°</td>
<td>-53±54°</td>
</tr>
<tr>
<td>ER</td>
<td>42±13°</td>
<td>87±16°</td>
<td>-45±11°</td>
</tr>
</tbody>
</table>

For concurrent validity, no differences were found between the groups (lay vs. pro), but a large bias was observed for all tasks on study population level (Table 1). Study population was therefore further assigned in two groups based on the observed bias (<16°: LOW bias group; >16°: LARGE bias group.). While the LOW group showed a bias [range] for ABD and FLEX of 1.0±3.1° [-2.5° to 5.9°] and -4.1±12.0° [-12.6° to 15.9°], the LARGE showed a bias of -68.9±27.5° and -93.2±36°, respectively. For both tasks, 5 subjects were assigned to LOW and 6 to LARGE group (data of 1 subject not available). For ER, all subjects were assigned in the LARGE group.

**DISCUSSION AND CONCLUSIONS**

Although every subject was able to complete the tasks, the usability study highlighted barriers limiting the subjects to successfully perform the exercises, primarily regarding communication and measurement accuracy (e.g. no error message, measurement stop during the tasks). As a consequence, two subjects reduced ROM of ABD to complete the measurements (due to a software crash at 180°), which resulted in a higher agreement between the two systems. No differences were found regarding preliminary knowhow (lays vs. pros), indicating that the measurement with the App is independent of user knowledge.

Concurrent validity was found to be high for 5 out of 11 subjects for FLEX and ABD. However, no subject was in the LOW bias group in both measurements, indicating that either FLEX or ABD but never both were measured correctly by the App. Further studies shall therefore evaluate validity with respect to smartphone placement on the upper arm. In addition, we recommend changing the order of the axis for Euler angle calculation in relation to the task so that primarily rotation (e.g. frontal plane for ABD) is always calculated first.

For ER, we found an averaged underestimation of 47.5±11.6% for the App. However, our Mocap was calibrated according to manufacturer’s recommendation for ABD and FLEX, and visual inspection of the video tapes suggests that the Mocap overestimated true ROM in ER. We therefore recommend reanalysing ER using another Mocap or at least another way of validation.

**CONFLICT OF INTEREST**

Two of the authors (RK and DB) were already involved in the App development. However, this has not influenced results and conclusions of this study.

**REFERENCES**

Introduction:
Functional Capacity Evaluations (FCE) are objective, standardized batteries of physical performance and functional measures that are frequently used to determine a person’s ability to perform work-related tasks [1]. An essential aspect of the FCE process is the interpretation of individual performance. Capacity outcomes of an FCE are documented via normative capacity data or a specific job demands analysis, but body mechanics and posture during an FCE lack the same guidance.

Normative upper extremity kinematic data for common FCE tasks can enhance understanding of typical movements during these tasks. Observation criteria is sparse and the available descriptions direct evaluators to classify each patient’s functional abilities based on their deviation from normal [2]. However, almost no description of normal is provided for various tasks. Normal movement must first be rigorously documented before deviations caused by injury or work intensity can be identified reliably.

The purpose of this project is to provide a comprehensive description of normative upper extremity movements strategies during select tasks of a Functional Capacity Evaluation.

Methods:
Thirty participants (15 males, 15 females; age = 23 (1.76) years) were recruited to for this study. Participants were excluded if they reported any upper extremity or back pain during functional tasks or any injuries to their upper extremities or back in the last six months. The study protocol was approved by the University of Waterloo Research Ethics Board and all participants provided written informed consent.

Participants performed three seated reaching and dexterity tasks that targeted upper extremity motions based on the WorkWell Systems FCE protocol [3]. The tasks were always performed in the same order; the repetitive reaching task (4 subtasks) was performed first, followed by the fingertip dexterity task (4 subtasks) and finally the hand and forearm dexterity task (2 subtasks). The repetitive reaching task required participants to move 30 marbles between bowls placed at their wingspan, while the Purdue Pegboard and Minnesota Manual Dexterity Test were used for the fingertip dexterity and hand and forearm dexterity tasks, respectively.

All movements were tracked 8 VICON MX20 (Vicon Motion Systems, Oxford, UK) optoelectronic infrared cameras and reflective markers were placed on the skin near anatomical landmarks consistent with ISB standards. Movement cycles were then defined within each trial and kinematic data were used to calculate time-varying 3-D relative joint angles of the torso with respect to the pelvis, humerus with respect to the torso, the forearm with respect to the humerus and the hand with respect to the forearm. Normative time series joint angle profiles were generated by ensemble averaging all participant curves to create a mean +/- one standard deviation profiles (Figure 1).

Results:
Kinematic profiles for each joint varied with task. During the repetitive reaching task, participants used an average absolute range of torso axial rotation range of motion of 40.1˚ (Figure 1a), while maximum humeral abduction and flexion reached 51.2˚ and 70.6˚, respectively. Elbow flexion consistently ranged from approximately 39˚ to 96˚ for the right arm, while both wrists maintained some level of ulnar deviation.

During the fingertip dexterity task, the range of motion for all joints was low. Torso angles remained consistent throughout each cycle. Each thoracohumeral angle had a range of motion of less than 15˚ for all subtasks, while mean humeral abduction decreased and both mean humeral flexion and humeral internal rotation increased from the unilateral to the bilateral tasks. Elbow flexion angle ranged from 61.2˚ and 77.0˚ for the right arm during all subtasks (Figure1b), while wrist motion was minimal.

Finally, during the hand and forearm dexterity placing task, torso axial rotation increased from 2.96˚ to 16.89˚ as participants moved from right to left. Mean humeral flexion and internal rotation also increased, while elbow flexion ranged from 45.7˚ to 106.1˚ and wrist angles were close to neutral. During the turning task, torso axial rotation reached peaks of 20˚ in both directions, while mean humeral axial rotation ranged from 20˚ to 45˚ of internal rotation (Figure 1c). Elbow...
flexion gradually increased from start to finish, and both wrists had an average extension angle of 9.8° and an ulnar deviation angle of 5.9° and 26.4° for right and left wrists, respectively.

Discussion:
The typical torso and upper limb kinematic profiles provided in this investigation is largest dataset of its kind to date. Clinicians and scientists will find the profiles useful because they provide a baseline to which motion can be compared to in order to better evaluate FCE performance. The kinematics of these tasks indicate their utility as evaluation tools for assessing the upper limb, specifically the shoulder and wrist. These reaching and dexterity tasks often required up to 60° of arm elevation, as well as high levels of humeral internal rotation. Also, for several subtasks in this investigation, participants used wrist postures at magnitudes that were a large portion of the available range of motion for the wrist [4]. Conversely, torso postures were almost always less than 30° from neutral and the elbow often remained within 60°-100° of flexion. This indicates the FCE tasks may not be as useful for evaluating these angles, but they should still closely monitored for potential compensations used by injured patients [5].

The establishment of a normative kinematic dataset for upper limb focused FCE tasks may improve evaluator’s return to work decision making. A normative dataset provides evaluators with the ability to compare their patient’s postures and movement strategies to typical healthy kinematics to enhance screening and interpretation of potential injury-mediated movement compensations or aberrations [6]. Used in conjunction with normative capacity data, these normative kinematic data improve resolution of individual ability to return to work and any limitations they may have [7]. This would improve treatment and return to work decisions. This work also supports the more ambitious future clinical goal of being able to identify people who are at risk of further injury or disability if returned to work too early.

References:

Figure 1: Mean kinematic profiles with +/- one standard deviation for a) torso +left/-right axial rotation during the right to left, right hand, repetitive reaching task, b) right elbow +flexion during the right hand fingertip dexterity task, and c) right humeral +internal/-external axial rotation during the turning task.
A novel virtual reality based approach towards the instrumented functional test of the shoulder

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INTRODUCTION
The main task of the shoulder is to position and orient the hand in a three-dimensional space. Dysfunctionalities in the shoulder, such as the very common fractures of the humerus, or the different forms of tendinopathy can cause severe impairments in the daily tasks of patients. During the rehabilitation period, most of these patients will develop compensatory movements that will allow them to perform their daily tasks with constraints due to their handicap.

Unfortunately, if the clinicians do not detect a lack of movement in a particular region of the reachable space, the avoidance of this region will hinder its return to normal function [1].

In the current clinical practice generally questionnaires and interviews are used to evaluate the pathologic limb [2]. These methods are subjective and could be completed with some objective measures through the use of instruments such as goniometers. Goniometers provide the maximal ranges of motion but lacks in three-dimensional evaluation.

Inertial sensors or smartphone with embedded sensors attached to the arm were used more recently as an effort to develop a more objective evaluation. While these systems provide an overall estimation of shoulder function and are validated for the evaluation of a wide array of pathologies [3], they do not provide detail about reachable space of the shoulder and regions in space where upper limbs have difficulty to reach.

In this work, we will follow the idea of an objective, fast and easy-to-use instrumented evaluation of the shoulder, while completing it with a more comprehensive three-dimensional evaluation of the reachable space. In this novel setup, the subject will be equipped with a set of PlayStation Move controllers (Sony Computer Entertainment) to track the motions of his trunk and hand, while reaching for a set of customized virtual targets provided in an immersive Virtual Reality scene. Two metrics evaluating the functional ability of the shoulder will be discussed.

METHODS
Motion tracking
Each subject was equipped with four PsMove controllers in order to record the motion of the shoulders, one hand and the contralateral hip. The three controllers on the trunk (shoulders and hip) were attached to a tightly fitting neoprene jacket, while the fourth controller was simply hold by the subject hand. The positions of these controllers were estimated optically in real-time by the means of a webcam tracking the respective light-bulbs of each controller. This information was combined with the data provided by the inertial sensors available inside each of the controllers to refine the position tracking, in particular in cases of optical occlusion. The open-source software Move.me was then modified to interface the data extracted by the PlayStation with a computer (via the network interface) to store the recorded data (markers trajectories, orientations and detection of light-bulbs on the image, as well as the raw data from the inertial units) and to interface the PlayStation with the virtual reality setup.

Virtual Reality
A simple virtual reality setup was developed to demonstrate the feasibility of the method: A cubic room with the avatar of the Ps-Move controllers floating in the space and following the movements of their real counterparts, as well as a green sphere (the target) floating in the air at the desired target position (figure 1). The display of the different target positions was regulated through a graphical interface on a control computer, allowing an operator to oversee the exercise through a reconstruction of the virtual world updated in real time. The same operator was also in charge to decide when the subject could not move any closer to the target position, and to display the next target.

The targets themselves were distributed along the surface of a sphere whose radius was the sum of the length of the arm and forearm of the subject, plus the distance between the wrist of the subject and the light-bulb of the controller in the hand. All the targets were placed in positions that a healthy subject could reach and the local density of these targets was defined with the help of a doctor in order to have more targets in
regions that are often impaired by the most common diseases of the upper limb.

Figure 1: Snapshot from the virtual reality, with legend.

**Studied Metrics**

We propose two simple metrics as an estimate of the reachable volume. First we used a volumetric score \( d \) as the average value of the Euclidian distances between the target points \( T(i) \) and the positions of the hand \( H(i) \) at the end of the reaching motion (as estimated by the operator during the recording).

Equation (1): 
\[
    d = \frac{1}{N} \sum_{k=1}^{N} ||H_k - T_k||
\]

Then, the ratio \( r \) of the score computed for the right arm (e.g. pathologic) against the left arm (e.g. sound arm) would offer an indication of the decrease in the reachable space.

Equation (2): 
\[
    r = \frac{|d_{right} - d_{left}|}{(d_{right} + d_{left})/2} \cdot 100
\]

To move to a more refined evaluation, the color-coded map of the target positions (with colours getting darker when the pathologic arm displays larger reaching errors than the healthy one) would help the therapists to identify the presence and position of a deficit in the reachable space.

**RESULTS**

By now three healthy subjects were recorded with the system. For each subject a total of 182 target points were obtained and their volumetric scores and ratios were computed and shown in Table 1.

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>d</th>
<th>r</th>
</tr>
</thead>
<tbody>
<tr>
<td>001 – Right arm</td>
<td>10.1 cm</td>
<td>25%</td>
</tr>
<tr>
<td>001 – Left arm</td>
<td>7.9 cm</td>
<td></td>
</tr>
<tr>
<td>002 – Right arm</td>
<td>9.3 cm</td>
<td>24%</td>
</tr>
<tr>
<td>002 – Left arm</td>
<td>7.2 cm</td>
<td></td>
</tr>
<tr>
<td>003 – Right arm</td>
<td>7.2 cm</td>
<td>11%</td>
</tr>
<tr>
<td>003 – Left arm</td>
<td>8.0 cm</td>
<td></td>
</tr>
</tbody>
</table>

The results show the technical feasibility of the recording method. Nonetheless the actual scores lack reliability due to the subjects feeling compelled to touch every target and moving their whole body (instead of moving their upper limb only) whenever the targets were slightly beyond their reaching capacity. Additionally, the evaluation of the distance between the hand and the floating objects, proved to be more challenging than expected and some visual cues will be added to the Virtual Reality scene to ease the process and reassure the subject that they are performing the task correctly.

**DISCUSSION AND CONCLUSIONS**

In this work, a novel approach for the functional assessment of shoulder joint was designed and tested. The main characteristics of this approach being (i) the ease of use of the system (about half an hour for a bilateral recording of 91 targets); (ii) the relatively small dimensions of the setup, allowing to use it in most medical cabinets; (iii) the quantitative and objective nature of the measurements; (iv) the possibility to record the performance of the subject in most of the reachable space without reducing it to a set of planes.

Following the preliminary tests, several minor adjustments will be necessary to guarantee the reliability of the measurements, such as the addition of more visual cues to help the subjects estimating the distance to the object within their reachable space, the redefinition of some targets in space that proved to be difficult to reach without masking some of the optical markers as well as a stronger form of penalization of the subject in the case of excessive movement of the trunk. Additionally a clinical validation will be prepared in order to establish the medical relevance and sensitivity of these new metrics. Measurements with more healthy subjects and patients with shoulder pathology is in progress.

**REFERENCES**


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Bilateral MRI findings in individuals with unilateral shoulder pain

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INTRODUCTION
Shoulder pain is a frequent musculoskeletal condition[1]. Bilateral deficits in different outcome measures have previously been reported in individuals with unilateral shoulder pain[2-6]. Previous studies have shown that scapular motion[4], humeral translation[5], abductor and scapular muscles performance[3,6] and mechanical sensitivity[2] are not different between symptomatic and asymptomatic shoulders in unilaterally involved patients. It is also interesting to know if soft tissue and/or bony alterations are present in these individuals to better understand the deficits already described in the literature. Therefore, the purpose of this study was to describe Magnetic Resonance Imaging (MRI) findings in both shoulders in individuals with unilateral shoulder pain.

METHODS
Thirty-six individuals with unilateral shoulder pain participated in this study. All had to have full active shoulder elevation range of motion (~150°) and pain for at least four weeks. Individuals with bilateral complaints, history of previous fractures and surgeries in the upper limbs, recurrent shoulder dislocations, body mass index > 28kg/cm² and neck pain were excluded. The Disabilities of the Arm, Shoulder and Hand (DASH) questionnaire was completed by all participants. MRI (Magnetom Essenza, Siemens®) with field strength of 1.5 Tesla and slice thickness of 3.5 to 4.0 mm was performed in all participants. Sequences in T1, T2 and proton density with slices in sagittal, coronal and axial planes were performed for both symptomatic and asymptomatic shoulders. A specialized musculoskeletal radiologist read all the exams.

RESULTS
Twenty men and sixteen women with a mean age of 41.2 (15.5 SD, 21-73) years were assessed. Nineteen (52.8%) individuals were involved in recreational overhead sports 1.5 (1.8 SD) times a week (10 with pain on the dominant side). The symptomatic side was the dominant side for 19 (52.8%) individuals (18 right-handed). The mean DASH score was 26.4/100 (18.56 SD, 1.7-79). Lower scores indicate better functional status. Surprisingly, MRIs revealed pathoanatomic structural alterations in all asymptomatic and symptomatic shoulders. Figure 1 shows alterations in both shoulders. The most common findings in the MRIs are presented in table 1.

![Figure 1: MRI findings of asymptomatic and symptomatic shoulders in a 73 year-old subject. A - T2-frontal plane; Arrows show a full-thickness tear of the supraspinatus and acromioclavicular osteoarthritis. B - T1-sagittal plane; circle shows fatty degeneration grade IV.](image)

<table>
<thead>
<tr>
<th>MRI finding</th>
<th>Symptomatic side (n=36)</th>
<th>Asymptomatic side (n=36)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tendinopathy</td>
<td>35 (97.2%)</td>
<td>35 (97.2%)</td>
</tr>
<tr>
<td>Partial tear</td>
<td>14 (38.8%)</td>
<td>11 (30.5%)</td>
</tr>
<tr>
<td>Full-thickness tear</td>
<td>2 (5.5%)</td>
<td>1 (2.8%)</td>
</tr>
<tr>
<td>Fatty infiltration</td>
<td>9 (25%)</td>
<td>7 (19.4%)</td>
</tr>
<tr>
<td>Labrum and biceps related lesions</td>
<td>21 (58.3%)</td>
<td>18 (50%)</td>
</tr>
<tr>
<td>Bursitis</td>
<td>22 (61.1%)</td>
<td>27 (75%)</td>
</tr>
<tr>
<td>Acromioclavicular osteoarthritis</td>
<td>30 (83.3%)</td>
<td>31 (86.1%)</td>
</tr>
<tr>
<td>Proximal humerus alterations</td>
<td>4 (11.1%)</td>
<td>7 (19.4%)</td>
</tr>
</tbody>
</table>
The most prevalent tendinopathy signs were observed in at least two tendons (supraspinatus [SS] and infraspinatus [IS]) or three (SS, IS and subscapularis [SB]) for both symptomatic and asymptomatic shoulders. Partial tears were observed with higher prevalence in isolation in the SS tendon or in combinations SS + IS or SS + SB. Fatty infiltration stage one or two was observed more frequently in three tendons (SS, IS and SB) followed by two (SS and IS) or only one (IS).

**DISCUSSION AND CONCLUSIONS**

Our findings were similar to previous studies that reported high prevalence of soft tissue alterations depicted in MRIs from asymptomatic shoulders[7-9]. MRI findings were previously described in asymptomatic shoulders of baseball[8] and tennis players[7]. These studies reported up to 86% of tendinopathy[8] and 40% of partial tears[7] in the dominant shoulders. Another study described more than 50% of superior or posterior labral tear in asymptomatic middle-aged patients[9]. Alterations in the asymptomatic shoulder seem to be common but often neglected. Therefore, it is important to be careful with misconceptions in seeking sources of pain using MRI. MRI findings may be more related to prognosis than treatment, since they seem to provide information that is more associated to progression rather than treatment of the dysfunction. Other models, such as a movement based system or symptoms modification procedures[10], should be considered to better explain shoulder pain. In conclusion, tissue alterations in the asymptomatic side of individuals with contralateral shoulder pain were as highly prevalent as in the symptomatic shoulders. These results suggest that MRI findings may be frequently overestimated and deserve less attention in the clinical decision-making process.

**REFERENCES**


**ACKNOWLEDGEMENTS**

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Experimental Shoulder Testing under in-vivo Anatomy and Physiology

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INTRODUCTION
Simulating arm abduction using an experimental shoulder simulator which matches physiological conditions would be of highest interest in shoulder biomechanics, e.g. to study the influence of shoulder anomalies on joint reaction force and muscular activity levels, to simulate rotator cuff tears to analyse different surgical treatment approaches, or to investigate primary stability of newly developed implants. For this purpose, we developed an experimental shoulder simulator [1]. However, previous conducted studies revealed that the increase of the joint reaction force while abduction is considerably decreased compared to in vivo measurements, and cranial subluxation was already observed at very low shear reaction force levels [2, 3, 7].

We therefore advanced the existing shoulder simulator by adapting anatomy and physiology to in vivo human shoulder condition by tilting the force action line of the rotator cuff muscles according to anatomical observations 15° caudally [4], and introduced primary antagonists: the Pectoralis Major and Latissimus Dorsi. The aim of this study was to compare compressive joint reaction force to literature data, to introduce a novel approach to study humeral head migration to detect cranial subluxation, and to analyse maximum shear joint reaction force of stable shoulders.

METHODS
For this purpose, a thiel-fixed humerus of a male donor was used according to medical-ethical guidelines and recommendations for working with cadaver material [5]. All soft tissues were removed with meticulous preservation of the whole enthesis of the following muscles: Infraspinatus (ISP), Teres Minor (TM), upper and lower Subscapularis (SSC), Supraspinatus (SSP), Deltoideus (DELT), Pectoralis Major (PM) and Latissimus Dorsi (LD). All muscle insertions were augmented with nylon strings and attached to the electro actuators by a rigid cable system (Figure 1). Muscles forces were controlled using unidirectional load cells (Model WMC, Interface). Tension load of primary abductors (SSP, DELT) were progressively increased in a balanced force ratio to abduct the humerus at a constant speed (6°/s). Rotator cuff muscles (ISP/TM, SSC) were constantly loaded throughout the whole range of motion with 22.5N each, and primary co-contractors (PM, LD) were controlled in accordance with electromyography activity patterns in relation to abduction angle [6]. An additional electro-actuator was used to control scapular rotation according to physiological shoulder rhythm (starting at 30° thoraco-humeral abduction angle in the ratio of 1:2 to humerus angle).

A size-matched, artificial glenoid (Mathys AG) was attached to a six-axis load cell to record and evaluate shear and compressive joint reaction forces. The humerus was additionally equipped with infrared markers of a motion capture system (Lukotronic, Steinbichler) to derive position data and calculate humeral head migration (Figure 1).

Figure 1: Experimental shoulder simulator with thiel-fixed cadaveric humerus and an additional weight of 2.5kg equipped with infrared markers.

For all conditions, five consecutive abduction motions were recorded in a frequency of 10Hz and averaged. Glenoid orientation was stepwise cranially tilted (5°) starting at +5° (caudal tilted) until subluxation. To describe shoulder joint stability, instability ratio was calculated by expressing shear in relation to compressive joint reaction force [3]. Position data of the motion capture system were transformed from global coordinates to glenoid fixed coordinates, taking into account the different glenoid orientations, to express humeral head migration along the glenoid
fossa. The same coordinate transformation was used for the joint reaction forces, so that the shear component always pointed along and the compressive component vertical to the glenoid fossa [2].

As humeral head motion is a combined rotation and translation on the glenoid fossa, the path of the instantaneous axis of rotation is subject to large sways. Therefore, migration of the predefined head centre was analysed and expressed in relation to vertical oriented glenoid (0°). All results were finally interpolated to obtain one data point at each integer abduction angle.

**RESULTS**

Compressive joint reaction force was in the range of literature data for all glenoid orientations (Figure 2) and significantly above previous data gathered with the same shoulder simulator for abduction angles above 35° [2, 3].

![Figure 2: Compressive joint reaction force for investigated glenoid orientations (coloured lines) in comparison to literature data [7] and previous studies with the same simulator [2, 3]. Error bars indicate 95% confidence interval.](image)

Based on humeral head migration (Figure 3), the -15° cranial tilted glenoid condition subluxated at 24° abduction angle.

![Figure 3: Cranio-caudal humeral head migration along glenoid fossa for investigated glenoid orientations (legend see Fig 2) with subluxation criterion (2 mm cranial migration, thin blue line) exceeded in -15° tilted glenoid condition at 24° abduction angle. Error bars indicate 95% confidence interval.](image)

For shear joint reaction force and thus the instability ratio, significantly higher maximum values were found as compared with previous data derived with the same simulator [2, 3].

![Figure 4: Shear joint reaction force (left) and instability ratio (right) for investigated glenoid orientations (legend see Fig 2). Note that the shoulder in -15° tilted glenoid condition (purple line) subluxated at 24°.](image)

**DISCUSSION AND CONCLUSIONS**

This study highlights the close to in-vivo measurements that can be performed with the advanced experimental shoulder simulator. By tilting the rotator cuff muscles (ISP/TM, SSC) according to anatomical findings and introduce the primary co-contractors (PM, LD), the compressive joint reaction force was found to be significantly increased and remains well within the range of in-vivo measurements [7]. By referencing humeral head migration to vertical oriented glenoid, we were able to quantify humeral head motion and thus to classify a cranial migration of 2 mm as subluxation. Although the subluxation was observed by two experienced clinicians (BM and SB) during the measurements, using the motion capture system allows to quantify subluxation and to assign a defined abduction angle. Furthermore, by optimizing simulators anatomy and physiology, we observed the cranial subluxation at a shear joint reaction force level of approximately 120N at an instability ratio of 0.44.

**FUNDING**

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**REFERENCES**

2. Moor et al, Clin Biomech (ahead of print), 2015
5. Dittmann et al, SAMW 2008
7. Baumgartner, Diss ETH No 18744, 2009
INTRODUCTION

The equilibrium of rotator cuff (RC) forces is disturbed in patients with RC tears, which may endanger shoulder stability and mobility. Knowledge on how RC tears affect shoulder kinematics in clinical patients may provide diagnostic information. Unfortunately kinematic signatures for RC tears in clinical patients are unknown yet. Moreover, patients with posterosuperior RC tears are underexposed in 3D motion analyses. Therefore, the in vivo impact of RC tear size and absence of supra- and infraspinatus forces on shoulder kinematics has not been demonstrated.

Differences in kinematics between subacromial pain syndrome (SAPS), isolated supraspinatus tear and massive posterosuperior RC tears are difficult to predict as stability and mobility require contradictory strategies. A cranial positioning of the glenoid (i.e. reduced scapular lateral rotation) or co-activation of shoulder adductors (i.e. increased scapular lateral rotation) may improve glenohumeral (GH) joint stability in massive posterosuperior RC tears, whereas mobility may require lengthening of the deltoid (i.e. increased scapular lateral rotation)1,3. Since co-activation of the teres major was previously observed in massive RC tears2, we hypothesized that the contribution of GH motion to the overall elevation is reduced in patients with a massive posterosuperior RC tear compared to GH motion in patients with SAPS and an isolated tear of the supraspinatus.

METHODS

Participants

Shoulder kinematics were prospectively evaluated in 109 consecutive patients with RC pathologies with 3D electromagnetic motion analysis at the Laboratory for Kinematics and Neuromechanics (Leiden University Medical Center, Leiden, the Netherlands) between April 2003 and October 2012. The clinical diagnosis was established after a thorough physical examination, antero-posterior shoulder radiography and magnetic resonance arthrography or computed tomography arthrography. In total, 34 patients with SAPS, 21 patients with an isolated full-thickness supraspinatus tear and 54 patients with a massive posterosuperior RC tear were included.

Shoulder kinematics

Shoulder kinematics of the affected shoulder were evaluated with the Flock of Birds (FoB), a 3D electromagnetic tracking system (Ascension Technology Inc., Milton, Vermont, USA) using seven wired sensors. Twenty-four bony landmarks were manually palpated and digitized according the recommendations by the International Society of Biomechanics (ISB)4. Patients were requested to perform abduction and forward flexion.

Figure 1: Glenohumeral motion (± standard error) in patients with SAPS (straight line), an isolated supraspinatus RC tear (dashed line) and a massive posterosuperior RC tear (small-dashed line).
Data processing
The bony landmarks were used to reconstruct a local Cartesian right-handed coordinate system for the thorax, clavicle, scapula, humerus and forearm according to the ISB recommendations\textsuperscript{4}. Humerothoracic motion, scapulothoracic (ST) motion and GH motion were calculated according to the appropriate Euler or Cardan angle sequence defined by Wu et al.\textsuperscript{4}. GH motion was described as: 1) GH plane of elevation, rotation around the scapular Y-axis, 2) GH elevation, negative rotation around the humeral X’-axis, and 3) internal GH rotation, positive rotation around the humeral Y’’-axis. For ST motion a fixed Cardan sequence (Y-X-Z) was applied: 1) protraction (i.e. internal rotation), positive rotation around the thoracic Y-axis, 2) lateral rotation, negative rotation around the scapular X’-axis, and 3) posterior tilt, positive rotation around the scapular Z’’-axis. We expressed humerothoracic elevation, ST lateral rotation and GH elevation as a positive motion.

![ABDUCTION](image)

Figure 2: Scapulothoracic motion (± standard error) in patients with SAPS (straight line), a supraspinatus RC tear (dashed line) and a massive posterosuperior RC tear (small-dashed line) during abduction.

RESULTS
Glenohumeral motion
GH elevation was significantly reduced in patients with a massive posterosuperior RC tear (figure 1). From 30° to 110° of abduction, GH elevation was raised by 3° (95%CI [1.5 – 5.4], p=0.001) to 16° (95%CI [10.5 – 21.2], p<0.001) in the SAPS group and 3° (95%CI [0.9 – 5.4], p=0.008) to 10° (95%CI [4.0 – 16.7], p=0.002) in the supraspinatus tear group. GH plane of elevation and rotation were indifferent between the pathologies. No differences in GH motion were observed between SAPS and supraspinatus RC tear patients. A comparable pattern was observed for GH elevation during forward flexion.

Scapulothoracic motion
Patients with a massive posterosuperior RC tear revealed significantly more ST lateral rotation (figure 2). From 30° to 110° of abduction, there was 2° (95%CI [0.5 – 3.4], p=0.010) to 11° (95%CI [6.5 – 15.2], p<0.001) and 2° (95%CI [-0.1 – 3.3], p=0.066) to 7° (95%CI [1.8 – 12.1], p=0.009) more lateral rotation in the massive posterosuperior RC tear group compared to the SAPS group and isolated supraspinatus tear group, respectively. A comparable pattern was observed for ST lateral rotation during forward flexion.

DISCUSSION AND CONCLUSIONS
Patients with a massive posterosuperior RC tear had less GH elevation and more ST lateral rotation compared to patients with SAPS or an isolated tear of the supraspinatus. GH elevation was indifferent between patients with an isolated supraspinatus tear and SAPS. These observations may indicate the important role of infraspinatus forces within the GH joint \textit{in vivo} for GH elevation in the presence of an isolated supraspinatus tear.

REFERENCES

ACKNOWLEDGEMENTS
This study was financially supported by the Dutch Arthritis Association (grant number 2013-1303).
The Effect of Humeral Elevation and Retroversion on Supraspinatus Subacromial Compression During a Simulated Reaching Task

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INTRODUCTION
Repetitive compression of the rotator cuff beneath the coracoacromial (CA) arch is theorized to be a primary mechanism in the pathogenesis of rotator cuff disease. Of the four rotator cuff muscles, the supraspinatus is especially at risk due to its location in the subacromial space. However, the kinematic mechanisms of rotator cuff disease are not well understood impairing the development and refinement of diagnostic and treatment strategies. The purpose of this study was to investigate the effect of humeral elevation on subacromial supraspinatus compression in individuals with and without shoulder pain. The effect of humeral retroversion was also investigated as a sensitivity analysis.

METHODS
Ten subjects with a history of shoulder pain (43±12 years) and nine subjects without a history of shoulder pain (35±7 years) were included in the study. Three-dimensional anatomical models of each subject’s humerus, scapula, supraspinatus, and CA ligament were reconstructed from magnetic resonance images obtained using a protocol designed to optimize tendon visualization. Regions of the supraspinatus muscle were defined as follows: (1) the footprint consisted of the portion of the tendon lateral to the articular margin; and (2) the tendon consisted of the footprint region and 1 cm medial.

Anatomical coordinate systems were constructed on the 3D models of the humerus and scapula based on published recommendations [1,2]. The coordinate system of the humerus was initially aligned to that of the scapula (i.e. neutral position). Due to the limited field of view of the MR scanner, the amount of subject-specific humeral retroversion could not be directly accounted for, because the epicondyles were not in view. Therefore, the humerus was internally rotated relative to the scapula from the neutral position by 57.2° to account for the average humeral retroversion (biceps method) [4]. Standardized glenohumeral kinematic data were then imposed on the models at 5° increments of humerothoracic elevation (Figure 1) according to average data from a previous study of asymptomatic subjects during a functional reaching task [3].

The proximity between the CA arch and supraspinatus tendon and footprint at each glenohumeral position was quantified using Euclidean distance. When contact occurred, the volume of the supraspinatus tendon that intersected with the CA arch was also quantified. Three-factor mixed-model ANOVAs were performed to investigate the effects of humerothoracic elevation (0°, 30°, 60°, 90°), retroversion adjustment from 57.2° (0°, ±20°, ±30°), and group (asymptomatic, symptomatic) on the minimum distances between the CA arch and the supraspinatus tendon and footprint. Including retroversion adjustment as a factor in the ANOVA served as a sensitivity analysis to investigate the impact of using the same retroversion magnitude for all subjects on the dependent variables. The volume of intersection was analyzed using Friedman’s non-parametric ANOVA due to non-normal data.

RESULTS
A significant two-factor interaction (humerothoracic elevation and retroversion adjustment) was observed for the minimum distance between the supraspinatus tendon and CA arch. In general, the supraspinatus tendon was closest to the CA arch at 30° humerothoracic elevation. However, the minimum distance at 30° humerothoracic elevation was not always significantly different from 0° or 60° (Figure 2).

Decreasing retroversion angle increased the minimal distance from the CA arch to the supraspinatus tendon.
at 60° and 90° (Figure 3). However, increasing retroversion had no effect on minimum distance across humerothoracic elevation.

The effects of elevation angle, retroversion angle, and group were complex for the footprint with the presence of a significant three-factor interaction. In general, the footprint was closest to the CA arch at 60° humerothoracic elevation. However, the minimum distance at 60° humerothoracic elevation was not always significantly different from 30° or 90°. At 0°, the minimum distance was 2.9 mm larger in the symptomatic group than asymptomatic group.

Intersections between the supraspinatus tendon and CA arch occurred in only 50% of subjects. Consequently these data were highly skewed. Freidman’s analysis across elevation angle was unable to determine significant elevation effects.

DISCUSSION AND CONCLUSIONS
During a simulated reaching task using average rotational motions, the supraspinatus is at greatest risk for subacromial compression between 0° and 60° humerothoracic elevation. The tendon clears the CA arch by 90° in most subjects however clearance remains small (<2 mm). These findings are meaningful because the current clinical assumption is that subacromial impingement risk occurs at higher elevations than demonstrated in this study.

A reduction in humeral retroversion angle increases minimal distance at higher angles of elevation. Therefore, decreased retroversion may potentially serve as a protective measure against mechanical rotator cuff compression at higher degrees of elevation. This is consistent with clinical thought that glenohumeral external rotation is beneficial to clear the rotator cuff under the acromion during overhead positioning.

Although humerothoracic elevation had no effect on volume intersections, there was a trend consistent with the minimal distance data. The highest volume intersection occurred at 30° with the lowest intersections occurring at 90° elevation.

This study has limitations that should be considered. First, standardized kinematic data were imposed on the anatomical models. Second, the humeral head was assumed to remain centered on the glenoid throughout the simulated motion. Thus subject specific kinematic deviations were not investigated. The primary objective of the study was to investigate the effect of glenohumeral elevation on subacromial proximity. Allowing for individual variability in glenohumeral kinematics and translations would have confounded the analysis. Therefore, these variables were controlled.

REFERENCES

ACKNOWLEDGEMENTS
This study was funded by the NIH/NICHD (R03, Ludewig), NIH/NIAMS (T32, Lawrence), and the Foundation for Physical Therapy (Lawrence, Staker).
Mechanical Internal Impingement of the Supraspinatus Tendon During a Simulated Reaching Task

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INTRODUCTION
Internal (or posterior) impingement occurs when the undersurface of the rotator cuff becomes entrapped between the glenoid and the humeral head. While this mechanism of rotator cuff injury was originally identified in overhead athletes during abduction and external rotation [1], it has also been observed during overhead elevation [2]. The magnitude of humeral retroversion is also theorized to impact the risk for posterior impingement, yet this relationship remains unclear. The purpose of this study was to examine the effects of humeral elevation and retroversion on the proximity of the supraspinatus tendon to the glenoid during a simulated reaching task.

METHODS
Eighteen subjects were included in the study. Nine subjects had no history of shoulder pain, and nine had either current or a prior history of anterolateral shoulder pain. Subject demographic data is presented in Table 1.

Table 1: Subject Demographics

<table>
<thead>
<tr>
<th></th>
<th>Asympt.</th>
<th>Sympt.</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
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<td>44 ± 12</td>
<td>0.07</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>175 ± 8</td>
<td>168 ± 10</td>
<td>0.16</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>77 ± 15</td>
<td>77 ± 17</td>
<td>0.96</td>
</tr>
<tr>
<td>Gender (males)</td>
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<td>0.64</td>
</tr>
<tr>
<td>Side tested (dominant)</td>
<td>7</td>
<td>9</td>
<td>0.47</td>
</tr>
</tbody>
</table>

Shoulder MRIs were acquired from each subject. Image files were imported into Mimics software where 3D anatomical models were created of the humerus, scapula, and supraspinatus. The supraspinatus tendon was defined as the footprint region and 1 cm medial.

Following construction of anatomical coordinate systems [3,4], the 3D anatomical models were rotated to a standardized magnitude of humeral retroversion based on the biceps groove method (57.2°) [5]. This was necessary because the field of view of the MR scanner was not large enough to visualize the epicondyles to define the lateral axis of the humeral coordinate system. The models of the supraspinatus tendon were then rigidly rotated through a simulated reaching task at 5° intervals of humerothoracic elevation using average data from a previous investigation of asymptomatic subjects (Figure 1) [6]. At each angle of humerothoracic elevation, the magnitude of retroversion was adjusted ±20° and ±30° to serve as a sensitivity analysis given the need to use a standardized magnitude across subjects.

RESULTS
The distance between the supraspinatus tendon and glenoid generally decreased at increasing angles of humerothoracic elevation (Figure 2).
A significant two-factor interaction (humerothoracic elevation and retroversion adjustment) was observed for the minimum distance between the supraspinatus tendon and glenoid (p<0.0001) (Figure 3).

Across all retroversion magnitudes, the average minimum distance was significantly smaller at 120° than at 90° (mean difference: 4.0-8.0 mm). The minimum distance at 150° was also significantly smaller than 90° for all retroversion magnitudes except when increased by 30° (mean difference: 5.8-9.1 mm). Only a reduction in retroversion significantly impacted the minimum distance between 120° and 150° humerothoracic elevation.

At 90° humerothoracic elevation, increasing or decreasing retroversion significantly reduced the minimum distance between the supraspinatus tendon and glenoid compared to the mean retroversion magnitudes except when increased by 30° (mean difference: 5.8-9.1 mm). At 120° humerothoracic, only an increase in retroversion significantly impacted the minimum distance. Specifically, the minimum distance was reduced by 1.1-1.2 mm compared to the mean retroversion.

Finally, retroversion had no effect on the minimum distance at 150° humerothoracic elevation.

DISCUSSION AND CONCLUSIONS

The risk for mechanical internal impingement is highest at angles above 90° humerothoracic elevation. Contact with the glenoid occurred during the simulated reaching task indicating entrapment of the undersurface of the supraspinatus tendon may occur in shoulder positions other than the abduction/external rotation position, which was originally identified as the primary position of risk [1]. This finding suggests internal impingement may be a mechanism of rotator cuff pathology in populations other than overhead athletes.

Changes in retroversion from a mean magnitude of 57.2° reduced the minimum distance and therefore increased mechanical impingement risk at 90°. This is likely because the anterior or posterior margins of the tendon are brought into closer proximity with the glenoid when retroversion is increased or decreased, respectively. At 150° humerothoracic elevation, however, altering the magnitude of retroversion did not have an effect on the minimum distance. Because the tendon was already in close proximity to the glenoid, the minimum distance metric may not be sensitive enough to detect changes in entrapment risk. Finite element modeling quantifying tissue strain may provide a more comprehensive assessment of supraspinatus risk in these overhead positions where the minimum distance is already minimized.

The results of this study help establish mechanical internal impingement as a potential mechanisms for rotator cuff pathology. Additionally, the results may inform ergonomic and exercise prescription to avoid positions that may place the supraspinatus tendon at an increased risk for entrapment against the glenoid.

REFERENCES

ACKNOWLEDGEMENTS

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Examining NSCA and ‘Powerlifting’ Bench Press Techniques on Upper Extremity Muscle Activity

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Introduction: Resistance training is the most common method to increase upper body strength in athletes and recreational lifters. The bench press is one of the most commonly prescribed upper body exercises [1], and several variations exist to maximize specific muscle activation or maximal load [2]. Two prominent methods for bench press exist: one emphasized by the US National Strength & Conditioning Association (NSCA), which uses a flat back on the bench, and a ‘powerlifting’ variant, using an arched back to increase lifting capacity. Previous research has focused primarily on the effects of bench angle or load instability [3, 4], but little research examines body postures that deviate from the standard NSCA protocol for bench press. This research aimed to determine how altering body position changes muscular activity during bench press resistance training.

Methods: 20 males (21.8 ± 1.7 years; 1.77 ± 0.07 m; 94.8 ± 19.5 kg) completed 6 submaximal bench press trials each: 3 using the NSCA protocol, and 3 using the powerlifting (ARCH) technique (Figure 1). All participants were experienced in both techniques at time of data collection, and included the bench press as a part of their regular training regimen at the time of collection. Prior to collection, participants were reminded of each of the techniques using a script. Trials were completed at 25, 50 and 75% of the participant’s one rep maximum (1RM). Average 1RM for these participants was 133.4 ± 21.9 kg. A spotter would assist the participant in lifting the bar from the rack when they were ready to complete the trial, but would otherwise not touch the barbell unless needed. Lifting cadence was as follows: two seconds eccentric loading (downward), no pause at the bottom, two seconds concentric loading. One repetition was completed during each lifting trial. Surface electromyography (EMG) was collected for middle deltoid (MDEL), pectoralis major – sternal (PECS) and clavicular (PECC) heads, latissimus dorsi (LATS), and triceps (TRIC) on the right arm at 1500Hz (Noraxon Telemyo 2400 T G2), linear enveloped and normalized to muscle-specific maximal outputs. Participants received a minimum of 2 minutes rest between trials to prevent fatigue.

Mean and peak values for each muscle were extracted. A 2-way repeated measures ANOVA (2 techniques * 3 normalized weights) was completed to determine the effect of technique and load on normalized EMG.

Results: Interactions between technique and %1RM occurred in mean LATS EMG (p=0.0247, Figure 2), with increased activation in the ARCH across all load levels. A main effect of technique emerged for mean and peak EMG for LATS and TRIC (p<0.0001, Figure 3), as well as mean MDEL EMG (p=0.0245).
In LATS and TRIC, activation was higher in the ARCH technique (Figure 3), while mean activity in MDEL was greater in the NSCA technique. Mean and peak EMG increased with increasing load for all muscles (p<0.05).

Discussion: Moving from the NSCA to the ARCH body position may increase internal rotation of the humerus during the negative phase of the lift. Increased elbow flexion may reduce force capacity for some muscles, increasing activation [5]. The interaction observed in LATS may have been generated by a training effect from the script (Figure 2). A main effect of load on mean and peak EMG in LATS, while still significant, was not as strong as any other muscle examined in this study, and was limited to the NCSA technique for the mean responses. Mean activity for LATS across all trials ranged between 9.7-11.8 %MVC, while peak EMG was between 26-33 %MVC. This relative stability of activation may be generated through the instruction rather than alterations in motion.

A main effect of lifting technique was present in both mean and peak EMG for LATS and TRIC, as well as MDEL mean activation. Using the ARCH technique resulted in increased activation for both LATS and TRIC, while decreasing MDEL activation. Pushes and pulls at shoulder height or higher lead to increased total activity compared to below the shoulder [5]. The posture adopted in the ARCH technique leads to decreased humeral forward flexion compared to the NSCA, and likely shifts some of the loading from the MDEL onto the larger

Conclusions: Using the ARCH lifting technique increases muscle activity in LATS and TRIC, but does not alter PECC/PECS activity. The utility of the ARCH technique to increase lifting capability is still unknown, but may provide increased capacity by shifting force generation requirements from the deltoids to LATS. Motion capture data was collected in conjunction with the EMG data reported. Future research will quantify changes in joint moments between techniques, and assess musculoskeletal geometric configuration changes and implications for force generation across observed postures.

References:
Electromyographic Investigation of Anterior and Posterior Regions of Supraspinatus: A Novel Protocol Based on Anatomical Insights

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INTRODUCTION

The supraspinatus is a complex muscle consisting of two anatomically distinct regions (Fig. 1): anterior and posterior, based on the lateral attachment of muscle fiber bundles onto the tendon [1,2].

Targeted studies have investigated functional and morphological aspects of these regions. In a real-time ultrasound study, Kim et al. [3] found significant differences in fiber bundle length changes between the anterior and posterior regions during active shoulder movements. The posterior region also has a higher percentage of Type II (fast) fibers [4]. Most recently, in a 3-D computer modeling study, Hermenogildo et al. [5] reported that the anterior and posterior regions are innervated by distinct primary nerve branches of the suprascapular nerve (Fig. 2). Innervation by distinct branches enables each region to independently modulate both the magnitude and the direction of force applied to the tendon.

The clinical implications of these anatomic and histologic findings are widespread as they challenge existing notions of the supraspinatus’ function and prompt evaluation of current rehabilitation strategies.

Although many EMG studies of the supraspinatus exist, only the anterior region has been investigated as per reported electrode placement; activation of the posterior region has not been investigated, precluding interpretation of its role in arm function.

The purpose of this study was to investigate the relative EMG activity of the regions of the supraspinatus during isometric actions for several shoulder abduction postures in the scapular plane.

METHODS

Twelve healthy participants with no previous shoulder or neck pathology from the university campus participated (5M/7F; age 32.5 ±12.2 years, range 22-61).

Two bipolar pairs of custom-made electrodes were inserted into the supraspinatus, each in the anterior and posterior regions. For the first six participants, electrodes were inserted separately. For the last six participants, to reduce needle insertions, the electrodes were paired and inserted.

To guide intramuscular electrode placement, external and internal landmarks were used. External landmarks included the acromion, spine of scapula, and clavicle (Fig. 3c). To estimate the anatomic
division of the anterior and posterior regions of supraspinatus, internal landmarks including the intramuscular tendon and the supraspinous fossa were identified with real-time ultrasound imaging.

**Figure 3.** Fine wire electrode placement. Ultrasound guided insertion into the anterior region (A) and posterior region (B). External landmarks (C). Tp = trapezius; SP = supraspinatus; *****intramuscular tendon.

All electrodes were inserted under the guidance of real-time ultrasound imaging (12 MHz linear array transducer) (Fig. 3a & b). The 3D model of muscle [2] and nerve [5] served as visual guides.

EMG activity levels from the anterior and posterior supraspinatus regions were recorded during an isometric exertion in three shoulder postures: 30°, 60° and 90° of thoracohumeral abduction in the scapular plane. Each participant held the arm position while holding a hand held weight set at 5% body weight.

EMG signals were recorded using a NORAXON EMG system and sampled at 3000 Hz. Data was transferred to a personal computer for processing using MATLAB™ (Version R2013b 8.2.0.701). Raw EMG data were processed via 4th order single pass Butterworth filter to produce a linear enveloped EMG response. EMG data for both recorded sites was rectified and smoothed via a root-mean square (RMS) filter with a moving average window of 500 ms.

The ratio of the activation levels for the anterior and posterior region was calculated for each arm position and participant. Data was then analyzed using SPSS statistical software. A Friedman test was conducted to evaluate differences in medians among all three arm positions. The Wilcoxon test was used for post hoc analysis. Bonferroni adjustment was made to account for multiple comparisons between ratios (P<0.017).

**RESULTS**
The ratio was influenced by arm position, $\chi^2 (2, n =12)=8.167, p = 0.17$. The median (IQR) ratio of anterior to posterior activation level for 30°, 60° and 90° of shoulder abduction were 0.59 (0.05 to 1.27), 0.47 (0.05 to 1.20) and 0.32 (0.07-0.70) respectively. There was a significant decrease in ratio between 60° and 90° positions ($Z=-2.353, P=0.016$).

**DISCUSSION AND CONCLUSIONS**
Perhaps one of the most important contributions of this study is that the activity of the posterior region has been investigated in an active exertion. This anatomically and functionally distinct region of the muscle has been overlooked in previous EMG studies, and thus recognition of its contribution to modulating the force transferred to the distal tendon has been absent. As no references exist for electrode placement in this region, we used detailed description from previous 3D modelling studies to establish best practice recording techniques that can be adopted in future studies.

The posterior region appears to play an increasingly important role at higher degrees of arm elevation. Limited research on the moment arms of rotator cuff muscular elements indicate that potential supraspinatus mechanical joint contributions vary with elevation [6]. Complementing this are documented changes in muscular activity in the anterior region [7] with elevated arm postures. However, these findings do not provide insight into the interplay between the anatomically distinct muscle regions when producing responsive forces to transfer to their common tendon. Further studies that consider different arm postures and exertions will help to refine descriptions of normal and pathological muscle regional contributions to arm function. This is especially germane to degenerative supraspinatus tendinopathy, which typically involves substantial morphological changes to the posterior region [8, 9].

**REFERENCES**

**ACKNOWLEDGEMENTS**
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Implication of Gluteus Medius Fatigue on Overhead Throwing Kinematics

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INTRODUCTION

It has become evident in the literature that pelvis and torso control plays a major role in dynamic human movement. With the pelvis and torso being a vital part of the lumbo-pelvic-hip complex (LPHC), optimal transfer of forces from the lower to upper extremity is required. For overhead movement it is imperative to have proximal stability in order to achieve distal mobility. The gluteal muscle group supplies the foundation of pelvis and torso stability over a planted leg thus resulting in efficient transfer of energy to the upper extremity.

It is known, that as a part of the kinetic chain, the gluteal muscle group affects LPHC stability as well as scapular stability. However, the true dynamic manner of the role of the gluteal muscle group and upper extremity kinetics during throwing has yet to be thoroughly examined. Furthermore, it is not known to what extent gluteal fatigue has on hip and shoulder passive range of motion (pROM). Therefore the purpose of this study was to induce gluteus medius fatigue on three consecutive days and examine changes in pROM as well as shoulder kinematics (humeral plane of elevation, elevation, and rotation) during overhead throwing. We hypothesized that after three successive days of gluteus medius fatigue there would be alterations in both pROM and throwing mechanics.

METHODS

Nineteen National Collegiate Athletic Association Division I female softball players (20.6 ± 1.9 years; 169.3 ± 9.7 cm; 73.2 ± 11.2 kg) participated. The University’s Institutional Review Board approved all testing protocols and informed consent was obtained. Data were collected on three consecutive days. Three days were chosen in attempt to mimic a typical competition season where a three game series is completed in three days.

The MotionMonitor™ (Innovative Sports Training, Chicago IL) synched with electromagnetic tracking system (Track Star, Ascension Technologies Inc., Burlington, VT) was used to collect data kinematic data, while a Baseline Digital Inclinometer (Medline Industries, Mundelein, Illinois) was used for bilateral hip and shoulder rotational pROM. Standard established methods for pROM were used. The examiner reported excellent intrarater reliability (ICC=0.90-0.98). Next, a series of nine electromagnetic sensors were attached to the following locations: (1) posterior/medial torso at C7, (2) posterior/medial pelvis at S1, (3) distal upper arm of the throwing shoulder (TS), (4) the flat, broad portion of acromion of the TS, (5) TS distal forearm, (6-7) bilateral distal/lateral upper leg, and (8-9) bilateral distal/lateral shank. A tenth sensor was attached to a stylus for digitization of the bony landmarks. Once the sensors were attached, the participant was given an unlimited time to warm-up. Once the warm-up was complete, participants were instructed to perform three maximal effort throws (60 ft/18.3 meters) to another player. Following the three throws, a bilateral gluteus medius fatiguing protocol was implemented. Repeated hip abduction, with the knee fully extended was performed to fatigue on a Biodex Isokinetic Dynamometer (Biodex Medical Systems, NY). Participants performed hip abduction throughout their full range of motion. Once the participant could no longer maintain full hip abduction for 3 consecutive trials or reported a rate of perceived exertion of a 20, on the Borg scale, fatigue was determined. After bilateral gluteal fatigue, participants performed three maximal effort throws followed by pROM measurements. Data from day one pre-fatigue and day three, post fatigue were selected for analysis. The throwing hip (TH) was defined as the ipsilateral hip to the TS and the non-throwing hip (nTH) was contralateral to the TS. Data describing the position and orientation of the electromagnetic sensors were collected at 100 Hz. Raw data were independently filtered along each global axis using a 4th order Butterworth filter with a cutoff frequency of 13.4 Hz. Euler angle decompositions were used to determine humeral orientation with respect to the thorax. Humeral orientation was determined as rotation about the y-axis of the humerus (plane of elevation), rotation about the z-axis (elevation), and rotation about the y-axis (axial rotation).
RESULTS
Means and standard deviations for shoulder kinematics and pROM are presented (Table 1 and Figures 1-2, respectively). Repeated measures ANOVAs were conducted to determine if statistically significant differences in pROM and shoulder kinematics, following three consecutive days of gluteal fatigue, were present (alpha level = 0.01, Bonferroni correction).

Table 1. Means and standard deviations of throwing kinematics pre and post fatiguing protocol expressed in degrees.

<table>
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<tr>
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<th>Ball Release</th>
<th>Max Internal Rotation</th>
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<tr>
<td>Plane of Elevation</td>
<td>2.0 ± 20.0</td>
<td>27.0 ± 41.1</td>
<td>23.6 ± 22.9</td>
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</tr>
<tr>
<td>POST</td>
<td>1.3 ± 22.4</td>
<td>17.7 ± 25.2</td>
<td>35.5 ± 22.0</td>
<td>50.8 ± 18.6</td>
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<th>Ball Release</th>
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<tbody>
<tr>
<td>POST</td>
<td>97.0 ± 14.4</td>
<td>102.7 ± 11.9</td>
<td>87.9 ± 9.9</td>
<td>75.6 ± 12.6</td>
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</table>

<table>
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<th>Max Elevation</th>
<th>Ball Release</th>
<th>Max Elevation</th>
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</thead>
<tbody>
<tr>
<td>POST</td>
<td>53.0 ± 38.4</td>
<td>114.3 ± 29.4</td>
<td>77.6 ± 24.1</td>
<td>34.4 ± 36.6</td>
</tr>
</tbody>
</table>

Figure 1. Pre and post intervention shoulder pROM (means and standard deviations). TS=throwing shoulder; IR=internal rotation; nTS=non-throwing shoulder; ER=external rotation.

DISCUSSION AND CONCLUSIONS
Bilateral gluteus medius fatigue did not cause alterations in hip and shoulder pROM or throwing kinematics. Because the body functions as a kinetic chain it was believed that fatiguing the musculature that helps provide LPHC stability would lead to kinematic alterations distally at the shoulder. The results indicate that shoulder mechanics during throwing do not change with gluteal fatigue. Since the gluteus medius functions as an internal rotator it was somewhat surprising that fatigue did not have a significant effect on hip pROM. The lack of change in pROM may be beneficial for clinicians working with collegiate softball player because it may indicate that changes in hip pROM are not drastic over a three-day period. Or it could the fact that fatigue and postactivation potentiation can coexist simultaneously in skeletal muscle thus resulting in greater neuroactivation and possibly not resulting in pROM or shoulder kinematic deficits.

There are several limitations of this study. The participants were all members of the same team. Thus, these results may not be generalizable across all softball players. The fatiguing protocol utilized may not be sport specific enough for softball, which is why no significant differences were observed. Future research should examine proximal-to-distal sequencing of the overhead throw following gluteal fatigue to see if efficient sequencing is maintained as well as explore the relationship between gluteal fatigue and postactivation potentiation.

REFERENCES
Spatial and load dependency of upper limb and shoulder muscle activity during submaximal exertions

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INTRODUCTION
Muscular fatigue and overexertion often precipitate the genesis of occupational injuries [1], particularly at the shoulder. While many work tasks occur at submaximal levels [2], the location of the task within the work envelope, and the load and direction of exertion may influence the overall muscular demand associated with each job task. The three-dimensional complexity intrinsic to the shoulder and upper limb precludes experimental examination of the muscular response at all possible hand locations within the work envelope. Quantitative description of multifactorial determinants of corporate muscular demand will provide evidence for task and workspace designs that reduce muscular demand during submaximal tasks. The objective of this work is to quantify shoulder and upper limb muscular demand during submaximal hand force exertions across hand location, hand force magnitude and force direction.

METHODS
Twenty right-hand dominant males (mean age: 21.6±2.7yrs) were evaluated. Muscle activation was recorded with surface electromyography (EMG) from 14 upper limb muscles: anterior, middle, posterior deltoid; biceps brachii; triceps brachii; clavicular and sternal regions of pectoralis major; supraspinatus; infraspinatus; latissimus dorsi; serratus anterior; upper, middle, lower trapezius. Measurements were recorded at 1500Hz (Telemyo 2400R T2; Noraxon, Scottsdale, AZ). Maximal voluntary contraction (MVC) myoelectric activity was also recorded for individual muscles using standard postures.

Subjects performed a series of submaximal exertions with the hand at five locations in the reach envelope. For each subject, the workspace was defined with the umbilicus as the origin, with +X-axis to the right, +Y-axis anterior, +Z-axis superior. All hand locations were located in a single plane parallel to the frontal plane, 30cm anterior to the origin (Fig. 1). A total of 120 hand exertion trials were performed, including one trial in each of the 5 hand locations for each combination of 4 occupationally-relevant, submaximal hand force exertion magnitudes (20N, 30N, 50N, 60N) in 6 directions (upward, downward, left, right, forward, backward). Each trial was 7sec, enabling the subject to ramp up to the desired force level. To reduce the effect of fatigue, 1min rest was given between trials. Hand force measurements were acquired with a 6-degree of freedom force transducer (MSA-6, AMTI, Watertown, MA) affixed to a robotic arm (Motoman HP50N, Motoman Inc., West Carrollton, OH).

The EMG data was processed by removing DC bias. Heart rate contamination was removed with a high pass 4th order Butterworth filter with a 30Hz frequency. EMG was linear enveloped, full wave rectified, and low pass filtered with a 4th order low pass Butterworth filter with 4Hz cut-off frequency. EMG signal was normalized by the corresponding muscle’s MVC. Average EMG representing each static hand force assessment was calculated over the 5-6sec window to avoid any ramping to the target force level. Total muscle activity, which is an assessment of overall muscular effort, was calculated by summing EMG from individual muscles.

$$\text{Weighted Total} = \sum_{i=1}^{14} \frac{MA_i \cdot PCSA_i}{\sum_{j=1}^{14} PCSA_j} [\text{Eqn.1}]$$

A weighted total muscle activity measurement was calculated, whereby each muscle activity (MA) measure was weighted by physiologic cross-
sectional area (PCSA) [3-5] and muscle specific tension ($\sigma=50.8\text{N/cm}^2$) [4], according to Eqn. 1.

**Statistical analysis**

One-way repeated measures ANOVA was used to separately examine the effect of hand position, hand exertion load and direction on weighted total muscle activity measurements. Two-way (magnitude, direction) repeated measures ANOVA was also used to examine the effect of direction and magnitude on muscle activity. Multiple linear regression with a forward stepwise method ($p=0.1$ to enter; $p=0.05$ to leave) was used to separately develop predictive equations for each exertion direction to predict weighted total muscle activity for variations in hand location and hand exertion load. All analyses were conducted with JMP software (SAS, Inc., Cary, NC).

**RESULTS**

Weighted total muscle activity was affected by hand location along the Z-axis (supero-inferior direction) ($p<0.001$), with greater muscle activity recorded for more superior hand locations. Weighted total muscle activity for hand locations along the X-axis (medio-lateral direction) was not different ($p=0.1297$). Muscle activity was dependent on hand exertion direction ($p<0.0001$), with downward exertions requiring the least total muscular effort and exertions directed upward requiring the largest effort (Fig. 2). Hand exertion load also affected muscle activity ($p<0.0001$); in each direction, weighted total muscle activity increased with increased hand exertion load.

Regression analyses indicate that weighted total muscle activity is dependent on hand exertion load and linear, quadratic, and interaction terms for horizontal and vertical hand positions within the 30cm anterior plane. Weighted total muscle activity across all hand locations demonstrated a linear increase for increases in hand exertion load. The coefficient of determination for regression equations ranged from $R^2 = 0.59-0.81$, indicating that up to 81% of the variability in measurements was captured by these equations.

**DISCUSSION AND CONCLUSIONS**

Muscular activity depends on hand position, hand exertion load and direction. These results are consistent with the increased effort needed to move the upper limb against gravity. This finding suggests that job tasks that require greater humeral elevation and exertions in upward directions may place workers at a greater risk for muscular overload. Prior work has identified the spatial and load dependencies for maximal [7] and a 40N submaximal hand exertion load [8-10]. The current study expands this work to quantify total muscular activity, which is a surrogate for overall demand, for variations in hand location, direction, and occupationally-relevant submaximal load. The differences identified in this study motivate continued development of regression equations to describe the muscular response to permutations of exertion direction (e.g. off-axis exertions), hand position, and loading conditions in the work envelope. These equations would permit evidence-driven designs of tasks and workstations to reduce muscular demand of workers and the risk for developing a shoulder injury for defined occupational force demands.

**REFERENCES**


**ACKNOWLEDGEMENTS**

NSERC Discovery Grant 311895-2011; Canada Foundation for Innovation; Ontario Research Fund.
Scapular Muscle Activity Pattern during Isokinetic Shoulder Flexion and Extension in Passive and Maximum Effort Conditions

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University Outpatient Clinic Potsdam, Potsdam, Germany

INTRODUCTION
It is assumed that frequent repetitive overhead movements under maximum load conditions in overhead sports or work put the shoulder at risk for injuries. Several investigations showed an association between shoulder pathologies and altered scapular muscle activity pattern [1, 2]. Those alterations seem to be dependent on the movement phase and the load applied [3]. However, previous work primarily focused on unloaded or submaximal loaded conditions. So far no study investigated scapular muscle activity in relation to the motion phase in neither pathologic nor healthy individuals under maximum effort conditions. Therefore, the purpose of this study was to investigate scapular muscle activity in regard to the motion phase during isokinetic shoulder flexion and extension in passive and maximum effort conditions in asymptomatic adults.

METHODS
The study was conducted in a cross-sectional design. Fourteen asymptomatic adults (m: 7, f: 7, 28±4 yrs; 1.75±0.12 m; 76±16 kg) volunteered to participate in this study. Shoulder flexion and extension of the dominant arm were tested in a standing position on an isokinetic dynamometer (Con-Trex, WS, Physiomed AG Germany). Simultaneously, muscle activity of the upper (UT) and lower trapezius (LT) and serratus anterior (SA) was assessed by a 3-lead surface EMG (myon RFTD-32, myon AG, Suisse). Participants performed isokinetic shoulder flexion and extension at 60°/s during a continuous passive motion (CPM), concentric (CON) and eccentric (ECC) loading. Testing in CPM was characterized by an isokinetic guided motion requiring no effort from the participant, whereas CON and ECC were performed with maximum effort. Range of motion (ROM) was defined between 20° and 180° shoulder flexion in the sagittal plane. After a warm-up participants performed five consecutive shoulder flexion and extension movements with prior familiarization for each isokinetic condition. EMG amplitudes were averaged over the five repetitions and root mean square (RMS) was calculated. ROM was divided into four equal motion phases for shoulder flexion (starting at 20° flexion) and extension (starting at 180° flexion). Muscle activity levels of each motion phase (RMS%) were normalized to RMS of the entire ROM. Data was analyzed descriptively (mean ± SD) and by a two-way repeated measures analysis of variances (ANOVA: α=0.05; post hoc Bonferroni correction α=0.017).

RESULTS
Absolute EMG values under CPM condition ranged from 75±47mV to 113±62mV for shoulder flexion and 31±13mV to 85±53mV for extension. For maximum effort conditions absolute values varied between 507±225mV and 667±299mV for flexion and between 161±56mV and 181±104mV for extension. Trapezius and serratus anterior muscles showed similar activity pattern. Passive shoulder flexion resulted in a linear increase of muscle activity, whereas the same movement under maximum effort resulted in an initial increase of muscle activity which was followed by a plateau (LT and SA) or decrease (UT) in the second half of the shoulder flexion movement (Table 1). For extension muscle activity decreased with the progression of the movement during the passive condition, whereas the activity levels during maximum effort initially decreased and started to increase again in the second half of the movement path (Table 2). The two-way repeated measures ANOVA revealed a significant interaction effect for condition and phase for both movement directions and all investigated muscles (p<0.05).

Figure 1: Isokinetic testing (60°/s) of shoulder flexion and extension (ROM: 20° - 180°) in sagittal plane. Movement phases (1. – 4.) are indicated by arm positions.
Table 1: Normalized scapular muscle activity (RMS%) during shoulder flexion for each of the four single movement phases. (LT= lower trapezius, UT= upper trapezius, SA= serratus anterior, *= significant differences (p<0.017).

<table>
<thead>
<tr>
<th></th>
<th>CPM</th>
<th>CON</th>
<th>ECC</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>43.2 ± 25.9</td>
<td>87.8 ± 26.6</td>
<td>65.5 ± 23.0</td>
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<tr>
<td>2</td>
<td>89.2 ± 27.7</td>
<td>93.4 ± 15.5</td>
<td>85.3 ± 17.1</td>
</tr>
<tr>
<td>3</td>
<td>109.7 ± 18.6</td>
<td>102.5 ± 15.8</td>
<td>106.7 ± 13.7</td>
</tr>
<tr>
<td>4</td>
<td>116.9 ± 32.2</td>
<td>105.8 ± 19.7</td>
<td>123.7 ± 19.0</td>
</tr>
</tbody>
</table>

Table 2: Normalized scapular muscle activity (RMS%) during shoulder extension for each of the four single movement phases. (LT= lower trapezius, UT= upper trapezius, SA= serratus anterior, *= significant differences (p<0.017).

<table>
<thead>
<tr>
<th></th>
<th>CPM</th>
<th>CON</th>
<th>ECC</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>132.5 ± 28.8</td>
<td>104.7 ± 44.9</td>
<td>124.2 ± 38.4</td>
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<tr>
<td>2</td>
<td>92.1 ± 18.8</td>
<td>67.4 ± 15.5</td>
<td>86.2 ± 24.9</td>
</tr>
<tr>
<td>3</td>
<td>82.0 ± 24.7</td>
<td>85.8 ± 20.9</td>
<td>76.0 ± 23.9</td>
</tr>
<tr>
<td>4</td>
<td>62.2 ± 21.2</td>
<td>113.2 ± 28.6</td>
<td>82.4 ± 31.1</td>
</tr>
</tbody>
</table>

DISCUSSION AND CONCLUSIONS

The aim of this study was to investigate scapular muscle activity pattern under varying load conditions and to further determine the influence of motion phase on activity levels.

It could be shown that scapular muscle activity is load and movement phase dependent as activity levels differed between passive and maximum effort condition and between type of work (CON, ECC). Regardless of the muscle, most differences in activity levels occurred in the beginning and the end of shoulder flexion and extension. The differences of scapular muscle activity pattern during distinguished isokinetic testing conditions indicate changing demands on the musculature resulting in higher activity levels and altered activity pattern throughout the movement.

When investigating scapular muscle activity in different populations or in the course of treatment evaluation it might be beneficial to evaluate single movement phases besides the activity level averaged over the whole movement.

REFERENCES

EVALUATION OF PASSIVELY INDUCED SHOULDER STRETCH REFLEx USING AN ISOKINETIC DYNAMOMETER IN MEN

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INTRODUCTION
The stretch shortening cycle (SSC) of muscle function was often biomechanically analysed by several investigators over the last decades. Strong evidence indicates that a SSC results in an enhancement of performance within the final phase of movement (concentric phase) when compared to isolated concentric action. However, so far, most investigations have focussed on the lower limbs [1].

First approaches suggested that the storage and release of elastic energy also cause a significant contribution in power enhancement within the upper body. Roach et al. [2] showed that “movements generated in adjacent, connected body segments, which can be transferred from joint to joint via a ‘kinetic chain’, including a rapid stretch of the shoulder internal rotators, enable elastic energy storage and release at the shoulder during overhand throwing.”

METHODS
Participants
Thirty healthy right-handed men (age … ± …, height … ± …, weight … ± …), with no history of shoulder musculoskeletal injury within the last twelve months or any shoulder weakness or discomfort during activity participated in this investigation. Before measurement, participants were informed of the experimental procedures and provided written informed consent.

Protocol
The study consisted of one familiarization and one test session. During test session electromyographic (EMG) sensors (Delsys, Bagnoli single differential) were attached to the pectoralis major muscle (pars clavicularis, pars sternocostalis and pars abdomialis), the deltoid muscle (anterior part, lateral part) and the latissimus dorsi muscle. Subjects were seated on an isokinetic dynamometer (IsoMed2000, D&R GmbH, Hemau, GER) in upright position with their left shoulder supported by a special backpad. The dominant right arm was abducted 90° and the elbow was flexed 90°. The resulting position was defined as the start setting (90/90-position, 0° external rotation) for the external shoulder rotation trials.

Table 1: Experimental conditions. The order of stretch conditions was randomized. (iMax: individual maximal stretch amplitude, MVC: maximum voluntary contraction)

To determine torque during a maximal voluntary contraction (MVC) each subject performed at least 3 trials of maximum isometric contraction in the 90/90-position with 2 minutes break between trials. The highest torque value during a steady plateau was used as MVC.

The individual maximum external stretch amplitude was determined at the same stretch velocity that was used during the test session (150°/s). Subjects were
instructed to keep their muscles in a relaxed status. Once they perceived any discomfort within the gelenohumeral joint or any surrounding muscle they were encouraged to actively contract their muscle to release a trigger which was set at 20% of the torque measured during MVC trials. Once the corresponding torque level was reached motion direction of the dynamometer was reversed. The averaged stretch amplitude of at least 10 trials was defined as individual maximal stretch amplitude (iMax).

External shoulder rotation stretches were then applied at a range of amplitudes and velocities and with different pre-activation levels according to the torque values of the MVC-test (Table 1).

Figure 2: External shoulder rotation stretch. Left: start setting (90/90-position), right: final stretch position.

For each condition a minimum of 10 trials was accomplished with a rest period of at least 10s between trials. The order of conditions was randomized. For verification of the actual joint position in comparison to the dynamometer position a kinematic model of the shoulder was used. Torque, actual external shoulder rotation, dynamometer position and EMG were sampled continuously. MVC trials were repeated at the end of the test session to ensure that the test protocol did not induce obvious signs of fatigue.

Data Collection and Analysis
EMG Surface EMG activity was recorded at a sampling frequency of 2000Hz via an A/D board (Nationl Instruments, Austin, Texas) and Templo 7.1 Software (Contemplas GmbH, Kempten, GER). Data were stored on a computer for further processing. Bipolar surface electrodes (Delsys Inc.; Boston, USA) with an interelectrode distance of 1cm were used. Before electrode placement the skin was shaved, abraded and cleaned with alcohol to minimise interelectrode resistance value. For EMG data processing MATLAB R2015b software (MathWorks, Natick, USA) was used. EMG signals were demeaned, full-wave rectified and bandpass zero-lag filtered (10Hz-450Hz) with a digital fourth-order Butterworth filter. To produce a representative EMG profile for each test condition, EMG signals from 10 trials were averaged. Stretch Reflex onset was then identified from the representative EMG profile as the first major deflection following the stimulus determined by visual inspection. Reflex amplitude was defined as Root-Mean-Square (RMS) EMG within a 20ms window from onset of the stretch response. Values were normalised to RMS EMG from the MVC trial during a 1s torque plateau.

Kinematic data Kinematic data were collected at 100Hz using four Basler A602f cameras (Basler AG, Ahrensburg, GER). Each subject had 10 passive reflective markers taped on the dominant arm and the torso. For kinematic data processing Vicon Motus 10 software (Vicon Motion System Ltd., Oxford, UK) was used.

Statistical Analysis
Person product moment correlation was used to examine relationships between reflex size, stretch amplitude and pre-activation level. Pre-post differences in MVC torque and reflex latencies were calculated using dependant sample t-tests. For all statistical analyses IBM SPSS Statistics 23 software (IBM Corporation, Armonk, USA) was used.

RESULTS, DISCUSSION AND CONCLUSIONS
Due to the fact that data collection took place in March 2016 by now only few results collected during pilot studies could be assessed. Pilot study data indicated that next to the pre-innervation level anticipation of stretch onset was of great importance for stretch reflex response.

Figure 3: Example of an averaged M.pectoralis major pars clavicularis response during full amplitude stretch condition.

REFERENCES
Experimental Analysis of the Subscapularis Force Response due to an External Load Impact

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INTRODUCTION
The knowledge about the forces transferred through tendons and ligaments during daily living or rehabilitation exercises is of interest in different areas of orthopedics. Analysis of the human shoulder forces is often done in a slow, controlled transfer of forces in order that stability of the model is maintained. Some literature is available to determine rotator cuff forces for sudden loading of the supraspinatus during car driving [1]. The force response in the co-contractile system of the subscapularis-infraspinatus mechanism would be of interest to determine loading at the rotator cuff also for other rotator cuff muscles.

Both muscles, subscapularis (SSC) and infraspinatus (ISP) influence each other during stabilising an internal/external rotation, which makes it difficult to work with computer models and predefined muscular algorithms.

SCOPE
The aim of the study was therefore a dynamic experimental analysis of subscapularis force reaction by inserting a sudden external loading on the arm.

METHODS
The muscle model
The measured muscle force characteristics has been evaluated by varying air pressure and contraction length of the artificial elements called « Fluidic Muscles » (Festo GmbH, Germany). For the SSC muscle, an air pressure of 1.3 bar has been chosen which resulted in a pre-tensional force of 30N.

Bull et al. [3] performed in-vivo measurements with two subjects by maintaining an isometric, resisted internal rotation to analyse needed subscapularis force. The tendon force was measured by arthroscopically insertable force probes AIFP’s. A correlation between an externally applied force and the subscapularis was determined (Figure 1) which was the basis to derive the muscle model on the basis of these in-vivo measurements.

The shoulder simulator
An experimental shoulder simulator has been developed including pneumatic muscles of Supraspinatus (SSP), Infraspinatus (ISP) and Subscapularis (SSC) representing the rotator cuff. Additionally, Latissimus (LAT) and Pectoralis (PECT) have been implemented to additionally stabilise the shoulder by applying a constant load. Fluidic muscle elements (Festo GmbH, Germany) have been used to simulate the physiologic behaviour of the muscles. By applying an air pressure, the element contracts and applies a tensional force. The compressive force at the glenoid was measured by a load cell. Additionally, all muscle forces were controlled with uniaxial load cells (Transmetra GmbH, Germany).

Subscapularis muscle controlled the angle of the humerus whereas infraspinatus on the contralateral
side was loaded with 30 N initially. The simulator just controlled the angle of rotation (resistance to internal rotation). The arm was mechanically fixed against abduction to isolate the movement just to humeral rotation.

**Testing procedure**

A sudden force was introduced by a falling weight of 5 kg and a drop height of 50 mm by a cable-pulley system in horizontal direction to the end of the lower arm, in a distance of 20 cm to the elbow. The force reaction on the glenoid load cell and on the Subscapularis was measured as well as the contraction length of the muscles and the resulting rotation angle. Data acquisition and steering/controlling the system was performed with labview realtime (National Instruments, US).

![Figure 3](image)

**Figure 3:** Biomechanical test setup using a flexed elbow where a horizontal impact force was introduced. The muscles were controlled in the way that the position was maintained by the subscapularis.

**RESULTS**

The sudden application of an external arm force by the falling weight resulted in a subscapularis force of 470 N to maintain the rotated shoulder posture. The orthogonal joint force component simultaneously increased to 500 N simultaneously. A maximum deflection of 20° was observed due to the flexibility of the muscle, until the controller regained the target value of zero degree position. The maximum force has been reached after 0.15 sec after force introduction.

It has been shown that the SSC muscle reaction forces increase rapidly and reaches the maximum value after 0.15 sec. The contraction of the SSC (SSC contraction) showed maximum values of 15 mm.

![Figure 4](image)

**Figure 4:** Subscapularis (F SSC) results in approx. 470 N whereas the glenoid force (F Glenoid) shows a maximum value of 500 N. Rotational arm angle along the humeral axis (Rot Angle) was deflected about 20°.

**DISCUSSION AND CONCLUSIONS**

The reaction force on a sudden loading of muscles to maintain the rotational position has been measured in present study. The force reaction of the SSC almost reaches the glenohumeral force and is therefore adequately high.

It is important to measure the loading response of muscles within a model due to the co-contractional effect of the infraspinatus, which influences the results of the loaded subscapularis behaviour.

Prevention of sudden loading after shoulder surgery is important to protect re-attached tendons postoperatively.

**LIMITATIONS**

A comparison of artificial muscles with in-vivo behaviour is only possible for quasistatic, isometric position. No data are available for dynamic, in-vivo force response.

**REFERENCES**

A framework for forward-dynamics simulation of the human shoulder

INTRODUCTION
A vast majority of the available biomechanical models of the human shoulder has been developed based on inverse dynamics, e.g. [1,2]. This imposes a number of limitations on their application. For instance, the glenohumeral joint is approximated as an ideal joint in an inverse-dynamics simulation. Therefore, the models fall short to predict the joint translations [3].

The different approaches developed to overcome the recurrent limitations of the models can be broadly divided in two categories. The first category tries to tailor an available inverse-dynamics model to a specific application, e.g. [3,4]. The second category aims to develop a framework allowing forward-dynamics simulation, e.g. [5,6]. Indeed, few studies have developed forward-dynamics simulations of the human body. In [5], dynamic optimization was used to develop a forward-dynamics model of the lower extremity. Dynamic optimization typically demands many times integration of the equations of motion. Given the computational expense incurred by the integrations, the method is impractical for common applications.

In this study, a framework for forward-dynamics simulation of the human shoulder is developed. In contrast with the dynamic optimization, the developed framework requires a single integration of the system equations. It is based on a joint application of a biomechanical model of the shoulder and a controller. The controller defines the muscle forces allowing the model to be simulated in forward dynamics. Different control scenarios are considered to investigate the model convergence in terms of accuracy and computational effort.

METHODS
Based on a given desired trajectory $q_d$, the controllers generate the associated muscle forces $(F + \bar{F})$ to steer the biomechanical model, as illustrated in Fig.1. Each of the blocks shown in Fig.1 will be now elaborated.

Biomechanical shoulder model
A model of the glenohumeral joint with three rotational degrees of freedom is derived. The scapula motion is considered by the scapulohumeral rhythm. All the 11 major muscles spanning the joint are included in the model as massless taut ropes. The paths taken by the muscles during the joint motion are defined using the geometrical wrapping algorithm presented in [2]. Using Lagrange’s equations, the equations of motion are derived:

$$ (\dot{H}_0 - M_0^{mg})[e_1] = \sum_{j=1}^{11} F_j(\rho_j \times n_j) \] [e_1] \quad \text{Eq.(1)} $$

where, $H_0$ and $M_0^{mg}$ are the angular momentum and the moment of the gravity force around the humeral head center, respectively. $[e_1]$ is the partial velocity matrix. $F_j$, $\rho_j$, and $n_j$ are the magnitude, the lever arm vector, and the direction vector associated with the $j$th muscle force.

Feedforward controller
The inverse system (if it exists) is always a candidate for the feedforward controller design [7]. Eq.(1), in compact form, can be written as

$$ D = [e_1][W]F \quad \text{Eq.(2)} $$

where $D_{3x1}$ is the left-hand side of Eq.(1). $W_{12x11}$ and $F_{11x1}$ are the moment arm matrix and the vector of muscle force magnitudes, respectively. By denoting $[e_1][W]$ with the quasi moment arm matrix $[B]$, Eq.(2) appears to be a linear algebraic equation. Therefore, if the matrix $[B]$ has full rank row, one can define $F$ associated with any given $q$ by solving Eq.(2). However, given the indeterminacy of Eq.(2), in order to arrive at a nontrivial solution for $F$, a static optimization routine is defined:

$$ \min \quad F^T[E]F $$

s.t. $D = [B]F \quad \text{Eq.(3)}$

where $[E]_{11x11}$ is a weight matrix and $F_{min}$ and $F_{max}$ are respectively the upper and lower bounds on the muscle force magnitudes. The cost function is the sum of squares of the muscle stresses [1,6]. The optimization routine defines $F$ such that it minimizes the cost, while satisfying the system dynamics (Eq.(2)) and the muscle-bound constraints.

Feedback controller
Having defined the muscle forces $F$ by the feedforward controller (Eq.(3)), the biomechanical shoulder model, given by Eq.(1), can be solved numerically for $q$. Ideally the resulted $q$ has to follow the predefined $q_d$. However, in practice the resulted $q$ starts off following $q_d$ reasonably well, but gradually loses accuracy as the time passes. More precisely, unless choosing a small enough stepsize for the simulation, accumulation of the successive errors due to the numerical integration causes the model response to drift away. However, the smaller the stepsize, the more computational effort is required. In order to ensure the model convergence for any reasonably large stepsize, a feedback controller is designed.

For the closed-loop system shown in Fig.1, the general form of Eq.(1) can be written as:

\[ 1 \]
\[ [M(q)] \ddot{q} + C(q, \dot{q}) + G(q) + [B](F + \bar{F}) = 0 \quad \text{Eq.(4)} \]

where \([M]\) is the inertia matrix, \(C\) is the vector of centrifugal force, \(G\) is the vector of gravity generalized force, and \(\bar{F}\) is the control input from the feedback controller. Given the simple nonlinear structure of Eq.(4), a feedback linearizing transformation is straightforward to derive [7]. Eq.(4) can be solved for \(\ddot{q}\)

\[ \ddot{q} = -[M(q)]^{-1}(C(q, \dot{q}) + G(q)) + [B](F + \bar{F}) \quad \text{Eq.(5)} \]

where the right hand side of Eq.(5) can be considered as \(V\), the new control input. This results in an equivalent linear system:

\[ \ddot{q} = V \quad \text{Eq.(6)} \]

We define the tracking error as \(\bar{q} = q - q_d\). Letting

\[ V = \ddot{q}_d - 2\lambda \dot{q} - \lambda^2 \ddot{q}, \lambda > 0 \quad \text{Eq.(7)} \]

results in an exponentially stable closed-loop dynamics. Having defined \(V\), the control input \(F\) can be achieved by substituting \(\ddot{q}\) from Eq.(6) in Eq.(4). Given the indeterminacy of Eq.(4), the same optimization routine as of Eq.(3) is performed to define \(F\).

RESULTS

A smooth motion consists of 150° abduction combined with 70° flexion and 35° external rotation is simulated. The motion is performed in 7.2 [s]. The Runge-Kutta-Fehlberg method, which combines a fourth and a fifth order Runge-Kutta scheme for error control is used to solve the differential equations [8]. The model response with the feedforward-only controller is shown in Fig.2 for three different stepsizes (0.00001, 0.0001, and 0.01). The response starts off following the given motion but it becomes far apart, except for \(T_s = 0.00001\). At this resolution the model response is almost indistinguishable from the given motion. This simulation took 6.5 [hrs] of CPU time on a 3.4 GHz processor with four cores.

The model response with the feedback+feedforward controller is shown in Fig.3 for two stepsizes (0.01 and 0.1). Comparing to \(T_s = 0.01\) with the feedforward-only controller, the tracking accuracy is phenomenally good while it takes roughly the same computational effort (135 [s]). The model response for \(T_s = 0.1\) provides an acceptable tracking performance. However in comparison with \(T_s = 0.00001\) of the feedforward-only controller, it requires 393 times less computational effort.

DISCUSSION AND CONCLUSIONS

A framework for forward-dynamics simulation of the human shoulder was presented. It consisted of a biomechanical model of the shoulder that was simulated in forward dynamics based on the muscle forces defined by a controller. Two different control scenarios were considered. The joint application of the feedback and feedback controller showed excellent tracking performance even for a reasonably large simulation stepsize (\(T_s = 0.1\)). However, the feedback-only controller could not exhibit the same order of accuracy even with 393 times more computational effort.

The developed forward-dynamics simulation provided a straightforward solution to the recurrent limitations of the available inverse-dynamics models. We will further develop the study by incorporating a more sophisticated shoulder model and accounting for the glenohumeral joint translations.

REFERENCES


ACKNOWLEDGEMENTS

This project was supported by the Swiss National Science Foundation [K-32K1-122512].

Figure 2: system response with feedforward-only controller, desired trajectory: solid line, system response: dash line.

Figure 3: system response with feedforward+feedback controller, desired trajectory: solid line, system response: dash line.
Simulation of forces in the glenohumeral joint with a multi body simulation model of the human shoulder

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INTRODUCTION:
The glenoid is a very complex joint in human body and stabilised by muscles and ligaments. It is therefore of great interest to know about the resulting muscle and contact forces. Knowledge about the force in the glenoid joint can give important information for medical treatment of shoulder injuries like humeral head fractures. Of particular interest is the influence of daily motions on implants.
An experimental test setup, already presented by the ZHAW Winterthur in the last conference of the International Shoulder Group, can provide such informations [1].
Additionally to an experimental test setup a computer-based simulation could be used for the validation of the model. For this purpose a multy body simulation (MBS) model based on this experimental setup could be beneficial for the research of the forces in the glenoid joint. The setup of the MBS virtual model is similar to the experimental model which allows a direct comparison of the results. (Figure 1)

![Figure 1: Test setup ZHAW (left) MBS model (right)](image)

METHOD:
The multi body simulation (MBS) of the human shoulder is modeled with the MBS software SIMPACK Version 9.9.1 (Dassault Systèmes, Vélizy-Villacoublay).
The model is composed of a scapula part which is linked via the glenoid joint to the humerus. The exact geometry of the humerus is based on CT data. It enables an abduction movement or a rotational movement of the upper arm. In order to specify the movement a nominal angle and duration is set, which is executed by the muscles. The muscles are represented by a force element, which works mainly as a cascade controller. The controller calculates the force of the muscle that is needed to reach the set angle. In addition the power is capped by the force value 'maximum force' to consider the limited power of a human muscle. While the muscle is in a relaxed state the force element will give back the basic tone of the muscle. All values for maximum force and basic tone are deposit in the related force element of each muscle. Table 1 shows the modeled muscles including their physiological cross sectional area (PCSA), resulting maximum forces ($F_{\text{max}}$) and basic tones [2]. The abduction movement is ensured by the rotation of the scapula and the elevation of the humerus while the scapula performs a third of the whole movement. Also M. supraspinatus and M. deltoid are implemented with a ratio of 1:3 (ssp : delt) to generate force needed.
In order to create realistic model conditions rotator cuff muscles, M. latissimus dorsi and M. pectoralis were added. These muscles generate a counter force for M. deltoid and M. supraspinatus and enable the rotational movement.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>PCSA</th>
<th>$F_{\text{max}}$</th>
<th>Basic tone</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deltoid</td>
<td>23.1</td>
<td>809.6</td>
<td>8.1</td>
</tr>
<tr>
<td>Supraspinatus</td>
<td>5.7</td>
<td>200.9</td>
<td>2.0</td>
</tr>
<tr>
<td>Subscapularis</td>
<td>15.9</td>
<td>556.2</td>
<td>5.6</td>
</tr>
<tr>
<td>Infraspinatus</td>
<td>5.1</td>
<td>178.5</td>
<td>1.8</td>
</tr>
<tr>
<td>Latissimus Dorsi</td>
<td>12.3</td>
<td>431.6</td>
<td>4.3</td>
</tr>
<tr>
<td>Pectoralis</td>
<td>13.6</td>
<td>476.4</td>
<td>4.8</td>
</tr>
</tbody>
</table>

Table 1: Modeled muscles with PCSA, $F_{\text{max}}$, basic tone [2]

RESULTS:
In the first simulation steps a validation of the MBS model was carried out by comparing forces with the measurement results of the experimental test setup [1] and in vivo data [3] (figure 2).
Considering the results of the abduction it becomes apparent that the values of the simulation have a similar trend as the results of the in vivo research published by the orthoload group. However there are differences between the maximum value of 564N compared to the in vivo results of 442N.
Compared to the results measured with the experimental test setup during the abduction show another variation in a drop of the muscle and joint force after reaching the maximum angle.
Additionally the forces of the M. deltoid as well as M. supraspinatus act like the results of the experimental setup of the ZHAW.
DISCUSSION:
With the MBS model of the human shoulder a powerful tool to investigate, resulting contact and muscles forces was developed. The comparison of calculated forces with the measurement results of the experimental test setup published by the ZHAW and in vivo data published by the ortholod group shows the validity of the MBS. In the next steps a fractured humerus treated with a bone plate or a nailing system modele das elastic bodies will be implemented in the MBS (fig. 6).

REFERENCES:
INTRODUCTION
It is widely believed that the most common cause of total hip replacement failure is aseptic loosening [1] due to osteolysis, which is primarily induced by polyethylene wear debris [2]. Consequently, hip simulators [3,4] have been designed to allow testing of implants in vitro to evaluate wear. Loosening and failure of shoulder implants has also been shown to be attributable to the wear of polymeric components in vivo [5,6]. However, prior to the Newcastle Shoulder Wear Simulator [7,8], no multi-station shoulder wear simulators existed capable of reproducing physiological ranges of motion and loading as seen in the human shoulder.

The Newcastle Shoulder Wear Simulator is a multi-station simulator capable of applying physiological motion in three axes with physiological loading. The simulator is fully programmable allowing it to reproduce activities of daily living [9], for example, lifting an object to head height, or drinking from a mug. The simulator allows different joint designs to be compared in vitro, and provides the implant designer with a tool to develop new prostheses prior to clinical use.

Previously, the simulator has been used to test JRI Orthopaedics Reverse VAIOS Shoulders [7,8]. In this study, 4 mm of sliding motion was added to the existing motions in a wear test of JRI Orthopaedics Total VAIOS Shoulders.

METHODS
The Newcastle Shoulder Wear Simulator is shown in Figure 1. Six JRI Orthopaedics Total VAIOS shoulder prostheses were wear tested in the simulator, five in the articulating wear stations with a sixth loaded soak control.

The prostheses were mounted in lubricant baths in diluted bovine serum with a protein content of 26 g/L. A 2 million cycle wear test was performed at 1Hz cycle frequency using gravimetric measurements to determine polymeric wear. Activities of daily living have been identified [9] and quantified in previous work [10] and ‘Mug to Mouth’ was selected as an activity of daily living for the wear test. Physiological loads were applied ranging from 180N to 250N, Figure 2 showing the measured physiological loading and the applied simulator loading. The physiological motion ranges were -16° to +12° in flexion-extension, -18° to -5° in abduction-adduction, and -42° to -17° in internal-external rotation with 4 mm of sliding motion. Polymeric wear debris was analyzed using a Nanosight LM10-HS, described elsewhere [7].
RESULTS
The wear results are shown in Figure 3 and over 2 million cycles the average polymeric components wear rates were 21.4 ± 4.5 mm³/10⁶ cycles. Figure 4 shows the results of this test of five Total shoulders with sliding motion and the previous test of five Reverse shoulders under the same conditions but without sliding, which was 13.3 ± 1.9 mm³/10⁶ cycles. Wear particles were sub-micron in size.

DISCUSSION
Previously, JRI Reverse V AIOS Shoulders were wear tested in the Newcastle Shoulder Wear Simulator [8]. In the absence of other shoulder simulators, the results were validated against hip simulators, as hip joints and reverse shoulder joints feature similar ball and cup geometry. The results showed excellent agreement with results from hip simulator studies of joints with the same materials when joint diameter, motion and loading differences were taken into consideration [8].

The wear particles from this and the previous study were both sub-micron in size as is observed with wear of other polyethylene implants [11]. Reverse shoulder joints take the form of a ball and socket and are essentially constrained by their geometry. Total shoulder joints generally have different radii for the glenoid and humeral components, and as such are not constrained in the same manner as Reverse Shoulders. The effect of this is to allow the joint to slide as well as rotate. 4 mm of sliding motion was added to the existing three axes of rotational motion in the simulator in order to test the Total shoulders.

The results of the wear test of Total shoulders with sliding motion were 21.4 ± 4.5 mm³/10⁶ cycles in comparison with 13.3 ± 1.9 mm³/10⁶ cycles in a wear test of Reverse shoulders without sliding. This 61% increase in wear rate can be attributed to the sliding motion.

CONCLUSION
The geometry of Total shoulder joints allows the possibility of sliding to occur in vivo. Adding sliding motion increases wear rate of the polymeric component. The increased wear rate of total shoulders due to sliding may challenge the clinical longevity of Total shoulders in comparison to Reverse Shoulders.

REFERENCES
Shoulder pain in individuals with spinal cord injury

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INTRODUCTION
Shoulder pain is a common health problem among individuals with chronic spinal cord injury (SCI). Spinal cord injuries may cause loss of sensory or motor control of lower limbs, trunk and upper limbs. Individuals with SCI mainly rely on their upper extremities during daily life (transfers, pressure reliefs, propelling, overhead reaching, etc.) and sporting activities. Shoulder pain has been shown to negatively affect mobility, participation and quality of life (Jensen, Hoffman et al. 2005).

The aetiology of shoulder pain in individuals with SCI is believed to be multifactorial. The loads applied on the shoulder and the capability to meet these demands were shown to play an important role in the development of shoulder pain (Figoni 2009). The exact pathways leading to shoulder pain remain, however, unclear.

Further research investigating population based-data and factors associated with shoulder pain is needed in order to develop targeted intervention programs. As such the aim of this study was twofold: (1) to determine the prevalence of shoulder pain and (2) to identify factors associated with shoulder pain (including sporting activities as well as mobility, independence, sociodemographic and socioeconomic factors, SCI characteristics, and health conditions) in a nationwide survey of persons living with SCI in Switzerland.

METHODS
This study used data collected through the 2012 community survey of the Swiss Spinal Cord Injury Cohort Study (Brinkhof, Fekete et al. 2016). In total, 1'549 eligible persons (Swiss residents with SCI aged 16 or older) participated (age: 52.3 ± 14.8 years; 29% female). Musculoskeletal shoulder pain over the past week (“yes”/“no”) as well as other predictor variables were assessed by self-report. The selected predictor variables included: sporting activities (never/past 4 weeks, 1-5x/past 4 weeks, ≥6x/past 4 weeks), independence in moving 10-100m (walking with orthoses or without aids, walking with aids and/or supervision, independent in manual wheelchair, electrical/partial assistance in manual wheelchair, total assistance), sex (male/female), age (16-30 years; 31-45 years; 46-60 years; 61-75 years; ≥76 years), education (compulsory school, vocational training, secondary education, university education), net equivalent income (low, medium, high), etiology (non-traumatic, traumatic), SCI severity (paraplegia incomplete, paraplegia complete, tetraplegia incomplete, tetraplegia complete), time since injury (0-5 years; 6-15 years;16-25 years; ≥26 years), spasticity (“yes”/“no”), and contractures (“yes”/“no”). Descriptive data analysis and multivariable logistic regression analysis were used, the latter providing adjusted odds ratios for shoulder pain for the selected predictor variables. The multivariable analysis was adjusted for item non-response by implementing a multiple imputation model and unit-nonresponse by applying inverse probability weighting derived from propensity scores.

RESULTS
The overall prevalence of shoulder pain was 35.8% (95% CI: 33.4-38.3). Multivariate regression analysis revealed higher odds of shoulder pain in females as compared to males (odds ratio; 95% CI: 1.89; 1.44-2.47) and when health conditions as spasticity (1.36; 1.00-1.85) and contractures (2.47; 1.91-3.19) were apparent. In addition, individuals with complete paraplegia (1.62; 1.13-2.32) or any tetraplegia (complete: 1.63; 1.01-2.62; incomplete: 1.82; 1.30-2.56) showed higher odds of shoulder pain as compared to those with incomplete paraplegia. No associations were found in the multivariable regression analysis with age, education, net equivalent income, etiology, and time spent with sporting activities.
DISCUSSION
This survey revealed a high prevalence of shoulder pain in individuals with SCI living in Switzerland. Higher odds of having shoulder pain were associated with females, individuals with tetraplegia and complete paraplegia, and with specific health conditions (contractures and spasticity) after controlling for all predictor variables.

CONCLUSION
Future longitudinal research is needed to define modifiable factors that are on the causal chain leading to the described associations.

REFERENCES

ACKNOWLEDGEMENTS
This study has been financed in the framework of the Swiss Spinal Cord Injury Cohort Study (SwiSCI, www.swisci.ch), supported by the Swiss Paraplegic Foundation. The members of the SwiSCI Steering Committee are: Xavier Jordan, Bertrand Léger (Clinique Romande de Réadaptation, Sion); Michael Baumberger, Hans Peter Gmünder (Swiss Paraplegic Center, Nottwil); Armin Curt, Martin Schubert (University Clinic Balgrist, Zürich); Margret Hund-Georgiadis, Kerstin Hug (REHAB Basel, Basel); Hans Georg Koch, (Swiss Paraplegic Association, Nottwil); Hardy Landolt (Representative of persons with SCI, Glarus); Nadja Münzel (Parahelp, Nottwil); Mirjam Brach, Gerold Stucki (Swiss Paraplegic Research, Nottwil); Martin Brinkhof, Christine Thyrian (SwiSCI Study Center at Swiss Paraplegic Research, Nottwil).
Proposal of a method to summarize quantitative motion analysis data of the upper limb for clinical applications

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INTRODUCTION
Quantitative analysis of human movement is a subject of increasing interest and dissemination. The clinical application of this discipline can be valuable for physicians, because it can help, for instance, in the process of performance-assessment, or in the evaluation of treatments. Often this analysis generates an abundance of data that can be very challenging to interpret and therefore they can be confusing. If the interpretation of data is difficult, clinicians tend not to use them because of their complexity, loosing the possibility to integrate quantitative and highly specific information in the standard medical evaluation. In motion analysis, one of the most ambitious goals is to generate a summary of parameters of clinical relevance, easily interpretable for clinicians. The aim of this study was to achieve this goal by proposing a simple graphical instrument. This is done through an example, i.e the study of shoulder and trunk compensatory movements of amputees while using a new multi-articulated prosthetic hand, a traditional myoelectric prosthesis and the patients’ own sound side.

METHODS
A group of 6 transradial amputees participated in the study (median age 47 years, range 35-65). Each subject underwent 3 motion analysis sessions: the first session using a tri-digital myoelectric hand (TD) (Fig 1a), which is the actual standard of care and allows the opposition grip only; the second session using an innovative prosthetic hand-wrist system, called “Michelangelo” (M – Ottobock, D) (Fig. 1b), which allows 7 grip; the third session using the Sound side (S). Within each session, subjects performed four standardized activities (each activity repeated 3 times). For the sake of brevity, only one activity is reported, i.e. Disk 1-2. Subjects were asked to seat in front of a table in a standardized resting posture. Two areas were marked on the table: ‘A1’, in front of the prosthetic hand; ‘A2’, in front of the subject trunk, at the maximum reachable distance, without moving the thorax. From the resting position they were asked to (I) pick up a standardized disk from A1, (II) move it to A2, (III) return to the resting position, (IV) pick up the disk from A2, (V) move it to A1, and (VI) return to the resting position. Data were captured through an optoelectronic system (Vicon). Markers setup is reported in Figure 2.

For each joint angle, a median curve over the repetitions was calculated, and its peak value saved for the group-level analysis. Group-level: the group median curve was extracted (Fig. 3) from the median curves of the 6 subjects, and a group median value was extracted for the time to complete the activity, and for the peak value of each joint angle [1]. For the data analysis, firstly the Friedman test and post-hoc comparisons “TD vs S”, “M vs S” and “TD vs M” were applied to identify statistically significant
differences among hands. Then, a second analysis was applied to select the clinically relevant differences among the statistically relevant. Three criteria were followed:
1) “Three-in-a-row” samples statistically different;
2) “Minimal required movement”, movement below the 30% of the maximum value of S were discarded;
3) “Minimum difference”, only samples with a minimum 30% variation between angles from two sessions were considered.

For each angle, the samples statistically different and clinically relevant were counted and then a colour-classification was carried out, following this code: Red when differences > 41% of the movement; Orange, 11÷40%; Yellow, 1÷10%; Green, 0%.

**RESULTS**

As shown in Figure 4 the use of TD leads to relevant alterations of shoulder kinematics compared to S. In Table 1 the results from the colour-classification are reported. Red cells highlights that there are more clinically significant differences between TD and S, compared to M and S. Differences between TD and M concentrate in the area of the subacromial impingement position, with substantial improvements in favour of M.

**DISCUSSION AND CONCLUSIONS**

The aim of this study was to show a new method for the interpretation of motion analysis outcomes, based on a graphical and easy-to-interpret analysis of data. A statistical analysis together with a clinical-specific assessment was used to assess data, and results were reported in a color-coded Table to permit a fast and intuitive interpretation. The Table shows that there are clear differences in motion of the scapula, humerus, girdle and thorax between TD, M and S. TD compared with S has large clinical differences with multiple angles. TD compared with M shows less clinical differences, but still far more than when M is compared with S, concluding that M can probably reduce the risk of injury for the rotator cuff, the back, thanks to the reduction of compensatory movements [2].

**REFERENCES**

**INTRODUCTION**

Cross country skiing became very popular recent years in Switzerland because of the Olympic success of Swiss athletes. However, training opportunities during summer are rare. An alternative training represents the roller skiing, similar to inline skating, where the sport devices consist of two wheels at both ends of a rail. Two poles are additionally used to apply a thrust force. Some literature is available about the loading of roller skis, in particular about the force distribution on fore- and rear foot during the kick off phase [1]. Only few investigation analyse the pole force for different techniques, most of them were focusing on the classic technique with sagittal plane motions [2].

Street and Frederick used a cable transducer to determine kick-off force of roller skis and ski-pole force in uphill skating by using a 100m long cable. The analysis of highly dynamic motion on an even terrain with a higher speed was therefore limited [3]. Almost no literature is available.

The analysis of acting forces on the ski pole is important to derive loading conditions at the shoulder joint, in particular to explain pathologic situations in case of overuse injuries.

The aim of the study was therefore the biomechanical analysis of acting force profiles on roller skis and on ski-poles for different skating techniques on an even terrain.

**METHODS**

Measurements were taken at constant speed under realistic training conditions for three athletes, which represented approximately 25 km/h. Strain was measured by strain gauges (HBM, Ger) on the bottom side of the skis, which were then used to calculate material stresses using the young’s modulus. The strain gauges were applied at the place where maximum bending stresses were expected.

The strain gauges on the ski pole compensated the occurring bending stresses in order that only compressive forces have been acquired.

Data acquisition was performed with a mobile wireless system, placed in a backpack of the athletes.

![Figure 2: Instrumented roller skis left (weasel®, Skiroll GmbH, Switzerland) and handle (right).](image)

![Figure 3: Applied skating techniques 2:1 and 1:1. The right ski and the right pole (in red) were equipped with strain gauges (=R). (Pole thrust for the 2-1-R-0 technique occurred parallel to the left kick-off, therefore = 0).](image)
Three different subjects have been used for the tests. A “semi-professional” skater was analysed in comparison with an “intermediate” and a “beginner”.

RESULTS
Generally, the pole force reached forces of approximately 220 N (28% BW) for the 2:1-technique and was therefore slightly higher in comparison to the 1:1-technique with around 150N (for the professional skater. For the intermediate and the beginner, the forces have been 20% and 40% lower respectively. Maximum stresses of 120 N were detected with the rear strain gauge on the skis which represents a loading of 2'000 N kick-off force or 250% BW (by measuring a speed of 25 km(h) for the professional. Interestingly, the pole forces show a tension during arm swing (shoulder extension) phase of 50 -70 N.

DISCUSSION AND CONCLUSIONS
The 1:1 technique results in significantly higher pole forces compared to the 2:1 technique during push-off (28% BW vs. 21% BW). A lower loading scenario could be therefore assumed for the 2:1 technique. Nevertheless, the shoulder speed during swinging phase (to re-position the arms in front of the body) needs to be analysed in detail because pole push-off occurs with every ski kick-off in 1:1 technique and results therefore in a higher stroke frequency. The measurement of the external force on the arms is the basis to determine inner forces at the shoulder joint by using 3D-musculoskeletal models such as Anybody or openSim. Acting muscle and joint forces are important to explain overloading in case of injury, because pathologies are still difficult to diagnose for this dynamic movement [4]. The present measurements of the pole force will be used therefore as an input parameter for computer simulations. The acting tension force on the poles during the arm swinging phase displays the highly dynamic motion to re-position the arm in front of the body. Such forces may lead to overtension of the ligamentous/tendinous structures in particular at the massively extended and internally rotated arm.

LIMITATIONS
No movement analysis was performed in that study which correlates arm motion with the occurrence of maximum pole thrust forces. The investigation is based on the examination of three subjects.

REFERENCES


ACKNOWLEDGEMENTS
We would like to thank Skiroll GmbH, Solothurn, Switzerland, for the technical support in the project.
INTRODUCTION
Shoulder diagnostic tests are primarily designed to place structures surrounding the joint under stress to elicit a sign or symptom that may help to identify the cause of a patient’s complaint. However laxity tests (anterior/posterior drawer, sulcus test), were developed to quantify, through subjective grades, the amount of glenohumeral translation during the test maneuver. The utility of these tests are typically assessed through reliability alone. No studies have examined the inter-examiner reliability of the translations being induced during the test maneuvers or the validity of the grades as they relate to the translations occurring during the exam. Therefore, the purpose of this study was; 1) to determine inter-rater, cross discipline reliability of translation magnitude for three joint laxity tests. 2) Describe the relationship between laxity test joint translations and subjective grades for anterior/posterior drawer and sulcus tests, considering both single tests, and scores combined across tests.

METHODS
Subjects
Kinematic data was collected from 11 volunteers with atraumatic, symptomatic shoulders. Institutional Review Board of the University of Minnesota approved the study protocol. Written informed consent was obtained from all participants prior to testing.

Instrumentation
Kinematic data was collected utilizing the Flock of Birds minbird electromagnetic system (EM) sensors (Ascension Technology, Shelburne Vermont) rigidly fixed to intra-cortical pins in the humerus and scapula. Data was processed using integrated MotionMonitor software (Innovative Sports Training, Inc. Chicago, IL.). Instrumentation static accuracy is reported to 1.8 mm and 0.5° (Ascension Technology Corporation).

Procedures
Intra-cortical pins were placed with fluoroscopic guidance into the humerus and scapula of subjects utilizing local anesthesia. Glenohumeral translations values were collected during the three laxity exam tests. The laxity tests were performed by a board certified shoulder surgeon (JB) and a physical therapist (PL) with expertise in clinical shoulder examination and biomechanics. Anterior and posterior drawer tests were performed with the examiner standing behind the subject, stabilizing the scapula with one hand while the other hand applied a compression force on the humerus into the glenoid followed by an anterior or posterior force along the glenoid. The surgeon assigned a grade of 0-3 for each test. The sulcus test is performed with the examiner stabilizing the scapula while an inferior force is applied to the humerus with the opposite hand. Grades of 1-3 were assigned by the surgeon for this test.

Data Reduction
Anatomical landmarks were palpated and coordinate systems assigned according to the International Society of Biomechanics recommended standard with the exception of use of the posterior AC joint rather than the posterior lateral acromion. Motion was described with Cardan and Euler angles and translations were described as displacement values of the center of the humeral head from the start of the test maneuver to the end of the maneuver.

Statistical Analysis
Intraclass correlation coefficients (Type 2,1) were calculated to assess the kinematic reliability of each test maneuver between the examiners. To assess the relationship of the examiners grade to the translation occurring during the exam, a simple linear regression was performed with the predictor variable as the single test grade and response variable as the single test translation. In addition, a separate regression analysis was performed utilizing a composite translation calculated as a root mean square (RMS) of the translations from each of the three laxity tests. This composite variable was set as the response variable with the predictor variable being the average subjective grade from the three laxity tests. Linearity of the results were checked and an a priori alpha level was set at 0.05
RESULTS

Inter-examiner translation reliability
During the test maneuvers, subjective reports of pain were less than 1.3/10 and attributed to either joint pain or pain associated with the intra-cortical pins. Reliability of the translations during the laxity tests is summarized in the Table 1. Anterior and sulcus test demonstrated good kinematic reliability (0.82 and 0.91 respectively).

<table>
<thead>
<tr>
<th>Test</th>
<th>Anterior Drawer</th>
<th>Posterior Drawer</th>
<th>Sulcus</th>
</tr>
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<tbody>
<tr>
<td>Examiner</td>
<td>JB PL</td>
<td>JB PL</td>
<td>JB PL</td>
</tr>
<tr>
<td>Trans. (mm)</td>
<td>3.4 4.2</td>
<td>-4.5 -1.5</td>
<td>-3.0 -3.1</td>
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<tr>
<td>ICC (CI)</td>
<td>0.82 (0.35-0.90)</td>
<td>na (0.55-0.94)</td>
<td>0.91</td>
</tr>
<tr>
<td>SEM (mm)</td>
<td>1.5</td>
<td>3.2</td>
<td>1.1</td>
</tr>
</tbody>
</table>

Trans, translation; ICC, intraclass correlation coefficient (2,1); CI, 95% confidence interval; na, not applicable (ICC calculation for posterior drawer not valid due to low between-subject variation); SEM, standard error of measurement.

Relationship of subjective grades to joint test kinematics
There was a poor association between the laxity grades for any single test and the magnitude of translation occurring during the test (Table 2). The average laxity grade from all three laxity tests and the composite translation calculated as an RMS value demonstrated a strong association ($r^2=0.74$, $p=0.0006$, Figure 1).

<table>
<thead>
<tr>
<th>Test</th>
<th>$r^2$</th>
<th>p-value</th>
<th>F-value</th>
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<tr>
<td>Anterior drawer</td>
<td>0.19</td>
<td>0.18</td>
<td>2.1</td>
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<tr>
<td>Posterior drawer</td>
<td>0.33</td>
<td>0.06</td>
<td>4.5</td>
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<tr>
<td>Sulcus</td>
<td>0.30</td>
<td>0.08</td>
<td>3.8</td>
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<tr>
<td>RMS of anterior drawer</td>
<td>0.75</td>
<td>0.0006</td>
<td>26.5</td>
</tr>
</tbody>
</table>

$F$, regression coefficient; RMS, root mean square of combined laxity tests.

DISCUSSION
This study differs considerably in how the reliability of the clinical exam was assessed. Previous approaches assess the reliability of the subjective grade with poor results and fail to determine if the examiners are inducing the same magnitude of translation for each subject. The results of this study suggest that kinematic reliability is possible for 2/3 tests designed to assess joint laxity.

Although a single test grade demonstrated poor association to the magnitude of translation occurring during the laxity test², the average grade and overall translation from the three laxity tests demonstrated strong association. Although clustering of tests results has been demonstrated in the past as a method to improve the utility of clinical exam techniques, this study provides a biomechanically based method for incorporating three tests to predict joint laxity. It also indicates that differences in the degree of joint laxity between individuals (microinstability) may be perceivable by the clinician when the grades from the laxity tests are taken as a whole.

REFERENCES
INTRODUCTION
In industrialised countries, between 7% and 30% of the population are affected by shoulder disorders. Numerous studies have also shown an increasing prevalence of shoulder disorders with age [1]. More than 20% of the elderly are affected by shoulder disorders, mainly due to rotator cuff diseases resulting in pain, lack of mobility or loss of strength [2]. The mobility of the upper extremity is crucial for activities of daily living and independent housekeeping. Shoulder disorders lead to restrictions in these activities. A patient assistive system (PAS) can be helpful for people with an impaired shoulder joint to compensate for reduced muscle forces and limited range of motion. Post-operative rehabilitation (e.g. after shoulder arthroplasty) and neuro-rehabilitation of stroke patients are further fields of application.

METHODS
Ease of use, low weight and an appealing design are essential for the acceptance of a PAS. Besides these, safety aspects are highly important for a device interacting with the human body. The system must be able to move the arm at least against gravity and the functional range of motion should be sufficient for activities of daily living. A range of motion of up to 120° of flexion and abduction is considered sufficient for typical daily activities [3].

RESULTS
The first version of an assistive system PAS Trio 1 was implemented as exoskeleton (Fig. 1). Additional load to the shoulder joint is prevented as rigid links bear the occurring forces. To enable a physiological motion, the scapula-humeral rhythm in the frontal plane was implemented. As a result the exoskeleton consists of three serial joints: the first replicates the rotation of the scapula, so that the axes of the second and third joint are aligned with the gleno-humeral joint. Each joint is driven by a DC motor and Bowden cables are used for force transmission. The exoskeleton is mounted on the upper back and the arm of the user is strapped to a brace. The system supports 120° of abduction, 120° of flexion and 50° of extension.

Figure 1: The first prototype PAS Trio 1 is an exoskeleton with 3 serial joints and a range of motion of up to 120° of flexion/abduction. It is adaptable to the patient and weighs 13kg. The design focus was set on the robustness of the system.

Figure 2: The prototype PAS Trio 2 was optimized for weight reduction, comfort and operability with EMG and voice control.
However, the PAS Trio 1 was bulky and not very user-friendly. Therefore, the prototype was optimized in terms of weight (7kg) and wearing comfort. For better usability the PAS Trio 2 (Fig. 2) can be controlled intuitively using surface electromyography (EMG) signals or alternatively with voice control. In EMG mode, the device detects the muscle activity of the deltoid muscle and adjusts the level of assistance accordingly.

As an intrinsic requirement of an exoskeleton, the mechanical axes have to be aligned with the anatomical axes of the shoulder joint. Misalignments can lead to harmful constraining forces in the human shoulder joint. Therefore, exoskeletons have to be adjustable to the user and the scapula-humeral rhythm of the shoulder needs to be taken into account. In consequence, a different approach was taken for the prototype PAS Mono 1 (Fig. 3) to further reduce weight, for easier adjustment to the patient and to minimize the risk of misalignment.

Figure 3: The prototype PAS Mono 1 is characterized by an appealing design, wearing comfort and usability at a range of motion of up to 120° of flexion/abduction.

The device is attached to a belt with a pivoting joint so that it can rotate around the vertical axis and supports up to 120° of flexion and abduction. The upper arm of the patient is strapped to a brace. The prototype PAS Mono 1 is driven by only one motor and controlled manually by joystick or surface EMG. The system is powered by battery and thus completely portable at a weight of only 2.5kg.

DISCUSSION
The PAS Trio systems are bulky and cumbersome to don and therefore not very suitable for daily life. The PAS Mono 1 is expected to achieve higher user acceptance because of the low weight and slender design.

For further improvement, the PAS Mono 1 will be evaluated by health professional for usability and suitability for therapy. The device will also be shown to potential patients to get feedback about their expectations and acceptance.

REFERENCES

ACKNOWLEDGEMENTS
The scientific work was financially supported by the ZHAW Zurich University of Applied Sciences.
## List of Participants

<table>
<thead>
<tr>
<th>Name</th>
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<td>Cristina Curreli</td>
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<td>Tobias Nef</td>
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<td>Jochen Springer</td>
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<td>Justin Staker</td>
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<td>Stefan van Drongelen</td>
<td>Orthopaedic University Hospital Friedrichsheim gGmbH</td>
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<td>Vassilios Vardaxis</td>
<td>Des Moines University</td>
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<td>Dirkjan Veeger</td>
<td>Afdeling Bewegingswetenschappen</td>
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<td>Monique Wochatz</td>
<td>University of Potsdam</td>
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<tr>
<td>Matthias A Zumstein</td>
<td>University Hospital of Berne</td>
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</table>
Programme at a Glance

Day 1: Thursday, July 14\textsuperscript{th} 2016
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur

<table>
<thead>
<tr>
<th>Time</th>
<th>Event</th>
<th>Location</th>
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<tbody>
<tr>
<td>13.30 - 18.00</td>
<td>OpenSim Workshop</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>18.00 - 19.30</td>
<td>Registration and Welcome Reception</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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Day 2: Friday, July 15\textsuperscript{th} 2016
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur

<table>
<thead>
<tr>
<th>Time</th>
<th>Event</th>
<th>Location</th>
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<tbody>
<tr>
<td>08.30 - 09.00</td>
<td>Registration and Coffee</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>09.00 - 09.15</td>
<td>Welcome</td>
<td>Baumgartner</td>
</tr>
<tr>
<td>09.15 - 10.10</td>
<td>Session 1 - Athletics and Sports (p. 14 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
</tr>
<tr>
<td>10.15 - 11.00</td>
<td>Keynote 1</td>
<td>PD Dr. med. Matthias A. Zumstein</td>
</tr>
<tr>
<td>11.45 - 13.00</td>
<td>Session 2 – Arthroplasty (p. 24 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
</tr>
<tr>
<td>13.00 - 14.00</td>
<td>Lunch</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>14.00 - 15.15</td>
<td>Session 3 - Scapular Kinematics (p. 34 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>15.45 – 17.00</td>
<td>Session 4 - Methods / Modelling (p. 44 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>17.00 - 18.30</td>
<td>Open Lab Session</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>19.00 - open end</td>
<td>Conference Dinner</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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Day 3: Saturday, July 16\textsuperscript{th} 2016
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur

<table>
<thead>
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<th>Time</th>
<th>Event</th>
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<tbody>
<tr>
<td>08.30 - 09.15</td>
<td>Keynote 2</td>
<td>Prof. Dr. Tobias Nef</td>
</tr>
<tr>
<td>09.15 - 10.15</td>
<td>Session 5 - Functional Assessment and Pain (p. 54 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>10.45 - 11.45</td>
<td>Session 6 - Rotator Cuff (p. 62 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>12.00 - 13.00</td>
<td>Lunch</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>13.00 - 14.15</td>
<td>Session 7 - Muscle (p. 70 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>14.15 - 15.00</td>
<td>ISG General Assembly</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>15.15 - 15.35</td>
<td>Session 8 - 2 Minutes Poster Presentations (p. 80 ff)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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<tr>
<td>15.45 - 16.45</td>
<td>Poster Session and Industry Exhibition with Wine &amp; Cheese</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
</tr>
<tr>
<td>16.45 - 17.00</td>
<td>Award Ceremony and Closing Remarks</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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Day 4: Sunday, July 17\textsuperscript{th} 2016
Location: ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur

<table>
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<th>Time</th>
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<tbody>
<tr>
<td>07.30 - 19.00</td>
<td>Expedition to the Swiss Alps, one-day trip (2 hrs walking)</td>
<td>ZHAW Winterthur, Technikumstrasse 71, TN E0.58, 8401 Winterthur</td>
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Wi-Fi access:       Network name:    ZHAW-event       Password:    58ZHAW74event