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In vivo measurement of shoulder joint loads during activities of daily living

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ABSTRACT

Until recently the contact loads acting in the glenohumeral joint have been calculated using musculoskeletal models or measured *in vitro*. Now, contact forces and moments are measured *in vivo* using telemeterized shoulder implants. Mean total contact forces from four patients during eight activities of daily living are reported here.

Lifting a coffee pot (1.5 kg) with straight arm caused an average force of 105.0%BW (%body weight) (range: 90–124.6%BW), while setting down the coffee pot in the same position led to higher forces of 122.9%BW on the average (105.3–153.4%BW). The highest joint contact forces were measured when the straight arm was abducted or elevated by 90° or more, with a weight in the hand. Lifting up 2 kg from a board up to head height caused a contact force of 98.3%BW (93–103.6%BW); again, setting it down on the board led to higher forces of 131.5%BW (118.8–144.1%BW). In contrast to previously calculated high loads, the contact force during passive holding of a 10 kg weight laterally was only 12.3%BW (9.2–17.9%BW), but when lifting it up to belt height it increased to 91.5%BW (87–95%BW).

The *moments transferred inside the joint at* our patients varied much more than did the forces both inter and intra-individually.

Our data suggest that patients with shoulder problems or during the first post-operative weeks after shoulder fractures or joint replacements should avoid certain activities encountered during daily living e.g. lifting or holding a weight with an outstretched arm. Some *energy-related optimization criteria* used in the literature for analytical musculoskeletal shoulder models must now be reconsidered.

1. Introduction

Knowledge of realistic loads acting in human joints is a prerequisite for improving design and fixation of joint implants, optimizing fracture fixations, and for improving physiotherapy. Furthermore, it allows the advising of patients to avoid any overloading of the joint or implant. Such data are also required to define realistic preclinical test procedures for implants, and can be the gold standard for validating analytical musculoskeletal models.

Until now, estimated loads acting in the glenohumeral joint were based on two or three-dimensional musculoskeletal models (Anglin et al., 2000; Van Der Helm, 1994) and kinematical analyses of everyday functional tasks (Van Andel et al., 2008; Veeger et al., 2006) or investigated by *in vitro* measurements (Karduna et al., 1998; Kummer et al., 1996). First validation of musculoskeletal models with *in vivo* measurements (Rasmussen et al., 2007) using *in vivo* measured data (Bergmann et al., 2007) showed quite realistic predictions for isolated movements like abduction. In other models, however, even this simple motion lead to variations of up to 30% (Favre et al., 2005; Poppen and Walker, 1978; Terrier et al., 2008; Van Der Helm, 1994).

Factors contributing to such discrepancies can include

- the large number of muscles involved and their unknown activation patterns,
- individual anatomical variations, and
- the change of muscle lever arms relative to the center of the glenohumeral joint or muscles wrapping around bones.

Furthermore, the function of the joints of the shoulder girdle is difficult to analyze and the accuracy of motion data is limited by skin movements (Veeger et al., 2003).

The aim of this study was to substantiate the first observations about high contact loads in the glenohumeral joint during certain activities of daily living (Bergmann et al., 2007), and to identify the factors contributing to such high loads by using instrumented joint implants.

Due to the enormous effort necessitated by such investigations, measurements are only possible in a small number of

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patients. This data cannot be generalized, but the measured loads are important as they are the only ones available. Despite the impossible hailing of statistics here, certain conclusions can still be drawn, and, comparing the measured with calculated loads will allow the improvement of analytical models of glenohumeral joint loading.

2. Methods

2.1. Instrumented implant

To measure the forces and moments transferred inside the glenohumeral joint, a clinically proven shoulder implant (BIOMODULAR, Biomet Deutschland) was modified. Six strain gages and a 9-channel telemetry are arranged in the hollow implant neck (Graichen et al., 2007). The telemetry is powered by an induction coil in the implant stem. The antenna at the tip of the stem transmits the signals. All electronics are safely sealed by electron beam welds. Detailed information about the implant is given elsewhere (Westerhoff et al., 2008).

2.2. Patients

Four patients with osteoarthritis of the shoulder participated in this study. Two were male, two were female. They were aged 62–80 years, abbreviated here as *S1R*, *S2R*, *S3L* and *S4R.* Their glenoids were in good condition and therefore not replaced. A deltoideo-pectoral approach was used. *The post-operative ROMs are given in* Table 1. Due to insufficient muscular strength, the 80-year-old patient *S4R* was not able to perform all tasks. Three patients obtained the implant on their right dominant side while patient *S3L* received it on her left, non-dominant side.

The Ethics Committee of our hospital approved implantation of the modified implant. Before surgery, the procedure was explained to the patients, and they gave their written consent to implantation of the modified shoulder implant, taking of the measurements, and publishing of their images.

2.3. Evaluation of data

All data are presented in the right-handed coordinate system for the right joint (Fig. 1, top right) as recommended by the International Society of Biomechanics (Wu et al., 2005). The data of *S3L* with the left-sided implant were mirrored. The coordinate system center is located in the center of the implant head. All forces

Table 1
Patients investigated.

Patient	*S1R*	*S2R*	*S3L*	*S4R*
Gender	Male	Male	Female	Female
Age	69	62	69	80
Weight (kg)	101	85	72	50
Implant head size (mm)	48	44	48	44
Replaced joint	Right	Right	Left	Right
Post-operative time (months)	3–11	2–9	3–7	Up to 3
Range of motion (ROM)				
Elevation: active (passive)	120° (130°)	160° (170°)	150° (160°)	110° (120°)
Abduction: active (passive)	75° (90°)	110° (120°)	130° (150°)	90° (100°)
External rotation: active (passive)	30° (40°)	30° (40°)	40° (50°)	30° (35°)
Internal rotation	90°	90°	90°	80°

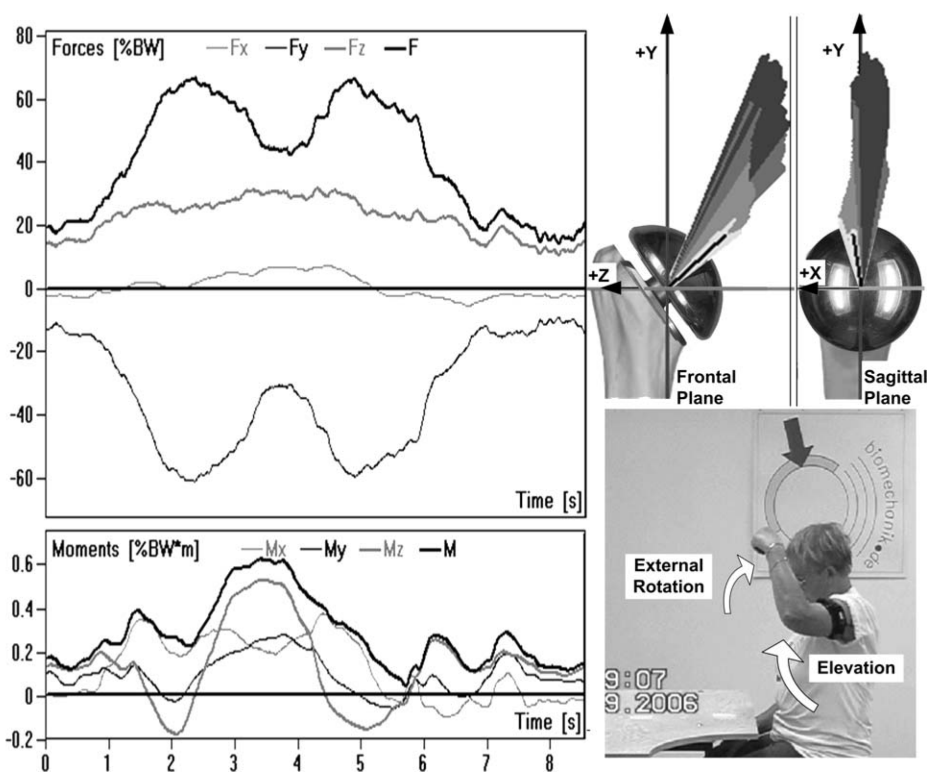

Fig. 1. Combing (Patient *S3L*; Activity 1). *Top left*: force components and resultant force vs. time. *Bottom left*: moments vs. time. *Top right*: force directions in two planes during whole movement (right joint). +*X*-axis points anteriorly, +*Y*-axis points cranially, +*Z*-axis points laterally in anatomical directions. For a better visibility, only resultant forces of more than 25% BW are displayed. The displayed prosthesis shows the exemplary replacement geometry and the center of the coordinate system in the center of the implants head. *Bottom right*: patient.

and moments, acting inside the joint, are measured relative to this point. The moments can be caused by two effects: (a) by friction between implant head and glenoid, or (b) an eccentric transfer of the joint force.

Measuring sessions on different days typically lasted 1 h with short breaks if the patients were tired. The patients were advised to repeat each activity 3–5 times, depending on their fitness, especially that of the older patients. Peak resultant force Fp and peak resultant moment Mp during each trial were automatically determined by customized software. These peak values from all trials of a single patient were averaged. Although the informative value of statistical tests with this small number of patients and repetitions has considerable restrictions, a difference between two activities of more than two std. dev. in the same patient comparing two activities was considered to be a meaningful difference. Comparisons between different patients are even more problematic because of the wide variations in anthropometric data, age and gender. To give a rough estimation of typical loads for a certain activity, data for an average patient (*AP*) were calculated from the averages of all investigated patients. All forces are given in percent body weight (%BW) and the moments in %BWm. Values in N or Nm are obtained by multiplication with 1% of the patient's body weight in Newton, e.g. by 7.36 for a BW of 75 kg.

2.4. Activities investigated

Investigated activities were chosen that are demanding for patients with shoulder implants, but nevertheless have to be performed frequently and are required for autonomous living:

1. *Combing* (Fig. 1): the patient moves a comb from the forehead along the middle of the head to the neck and then back. No other restrictions were given.
2. *Two-handed steering* (Fig. 2): in a sitting position the movement starts with the

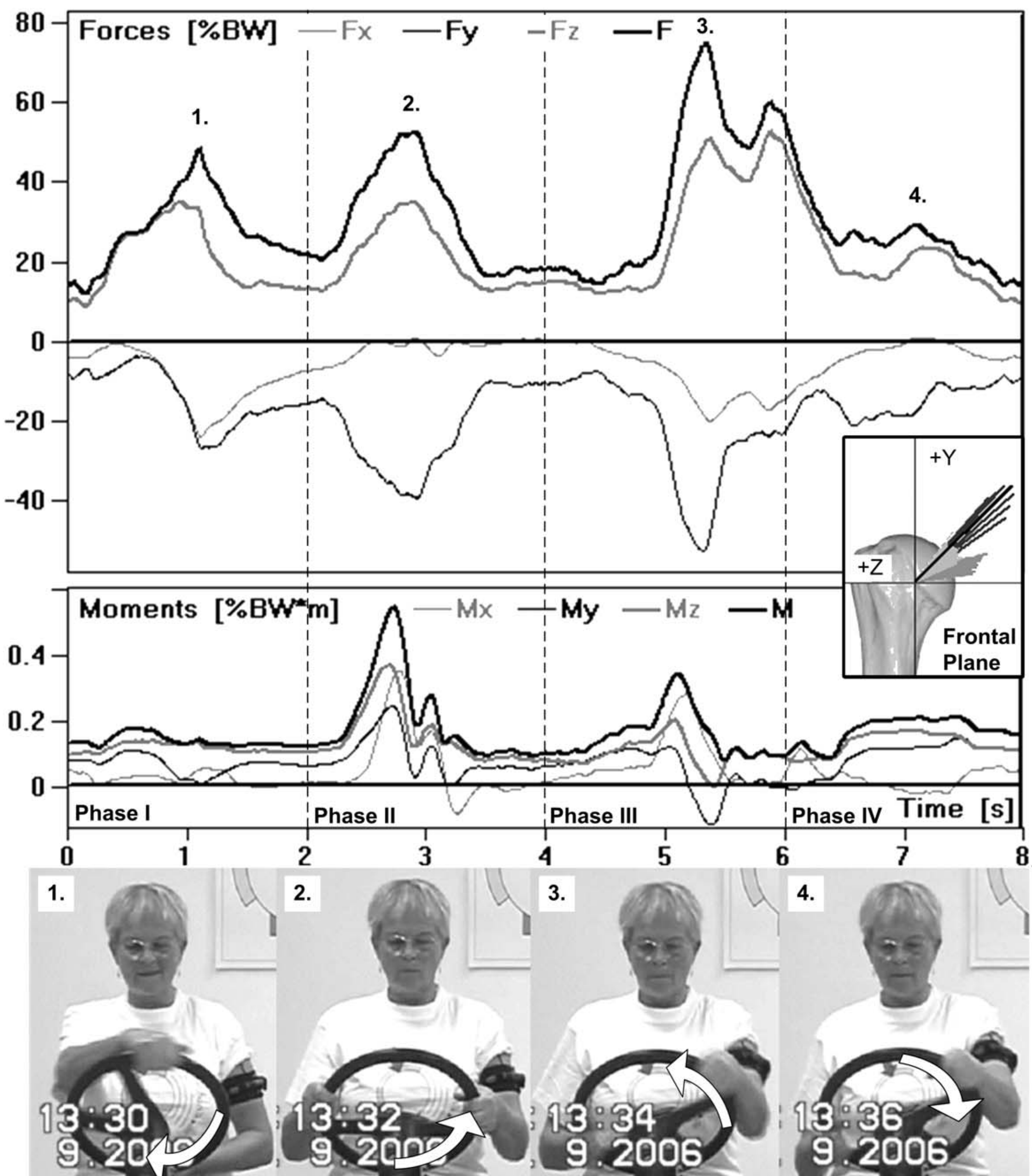

Fig. 2. Steering two-handed (Patient *S3L*; Activity 2). *Top*: force components and resultant force vs. time. *Middle*: moments vs. time. *Insert in diagrams*: force directions in frontal plane during whole motion. Movement starts with hands in horizontal positions. During the first 90° turn of the wheel, the hand on the implant side is lowered (phase 1) and moved back (phase 2). Then a 90° turn to the other side (phase 3) and back (phase 4) is performed.

hands in horizontal position. During the first 90° turn of the wheel the hand on the implant side is lowered (phase 1) and moved back (phase 2). Then a 90° turn to the other side (phase 3) and back (phase 4) is performed. The resistance was adjusted to 7 Nm with a friction clutch. Sitting position and wheel inclination were adjusted according to the patient's preference.

3. *One-handed steering:* except for *S4R*, the patients also performed the steering exercise as described before, using only the hand on the implant side.
4. *Nailing* (Fig. 3): facing a wall, the patient holds a 500 g hammer with a horizontal upper arm and an upright forearm, and then hammers a nail partially into a wooden board.
5. *Holding and lifting* 10 kg *laterally* (Figs. 4 and 5): the patient passively holds a crate with bottles of water (10 kg) using one hand laterally (Fig. 4; 0–0.7 s). Then, the crate is lifted upwards vertically by flexing the elbow (0.7–2.5 s) and lowered again (2.5–5 s). *S1R* and *S2R* are also performed this task at a self-chosen greater speed.
6. *Lifting a coffee pot* (Fig. 6): the patient sits at a table and lifts a coffee pot (1.5 kg), placed close to the upper body, stretches out his arm, and sets the coffee pot down onto the table. Then this sequence is reversed.
7. *Board at belt height* (Fig. 7): the patient is standing beside a table where a weight of 2 kg was placed. Then the weight is picked up, put on a board at belt height, located in a distance of 30 cm, it is then taken back, and set down onto the table again.
8. *Board at head height:* patients *S2R* and *S3L* also performed the board task with 2 kg when the board was at head height, and the arm had to be stretched out to set down the weight.

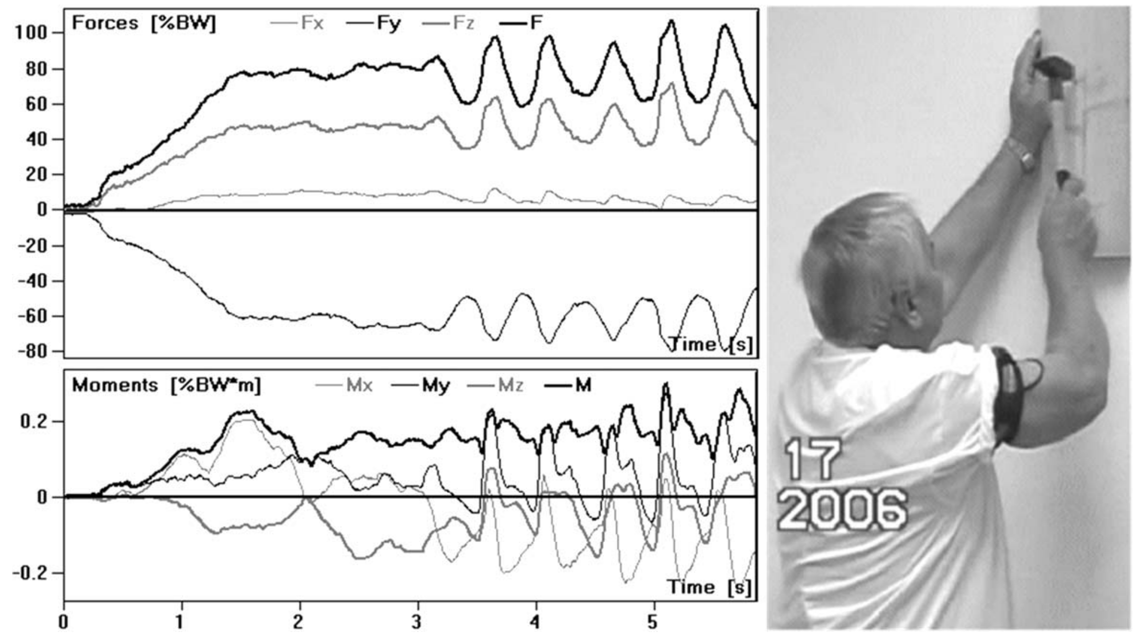

Fig. 3. Nailing above the head (Patient *S2R*; Activity 4). *Right*: patient image. *Top left*: force components and resultant force vs. time. *Bottom left*: moments vs. time. The arm hangs down (0–0.3 s), the hammer is lifted (1.6 s) and held (3 s), nailing with horizontal upper arm and upright lower arm is done with increasing force (6 s).

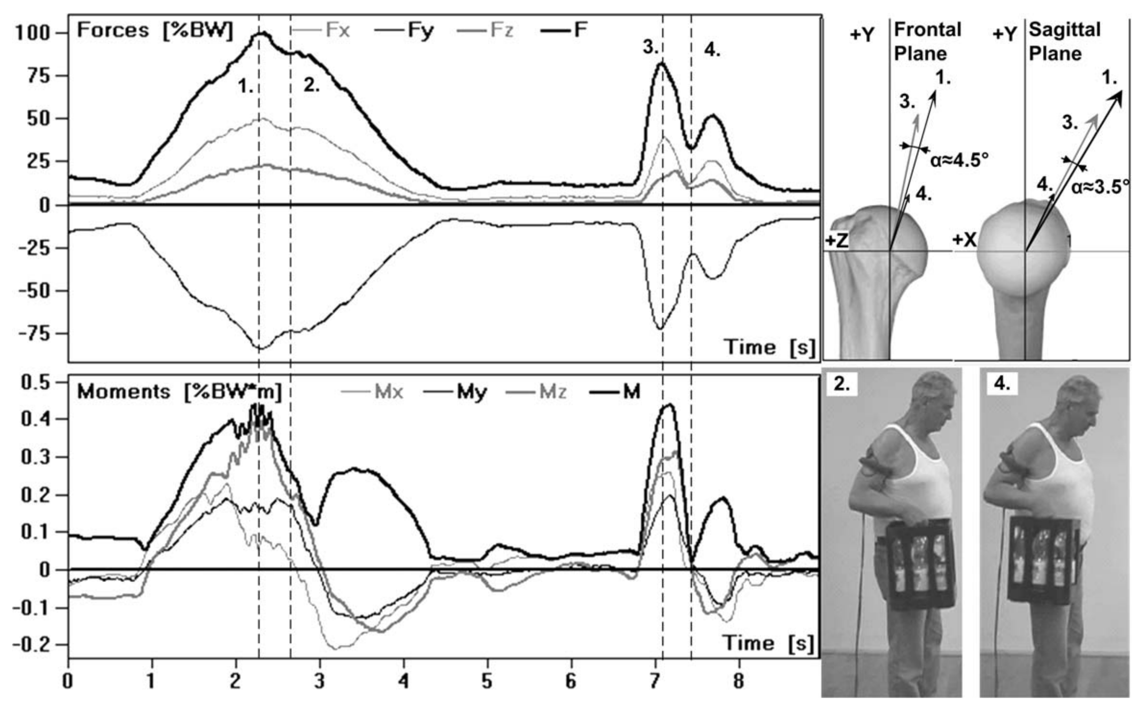

Fig. 4. Holding and lifting 10 kg (Patient *S1R*; Activity 5). *Top left*: force components and resultant force vs. time. *Bottom left*: moments vs. time. *Top right*: force directions in frontal and sagittal plane during whole movement. *Bottom right:* points of highest elevation for slow and fast speed. The weight (10 kg) hangs down passively (0–0.7 s), is slowly lifted by 39.2 cm (2.3 s), and lowered (4.9 s), then is lifted faster by 40.9 cm (7 s), and lowered (8 s). Peak forces (1. and 3.) occur before the weight reaches its highest position (2. and 4.).

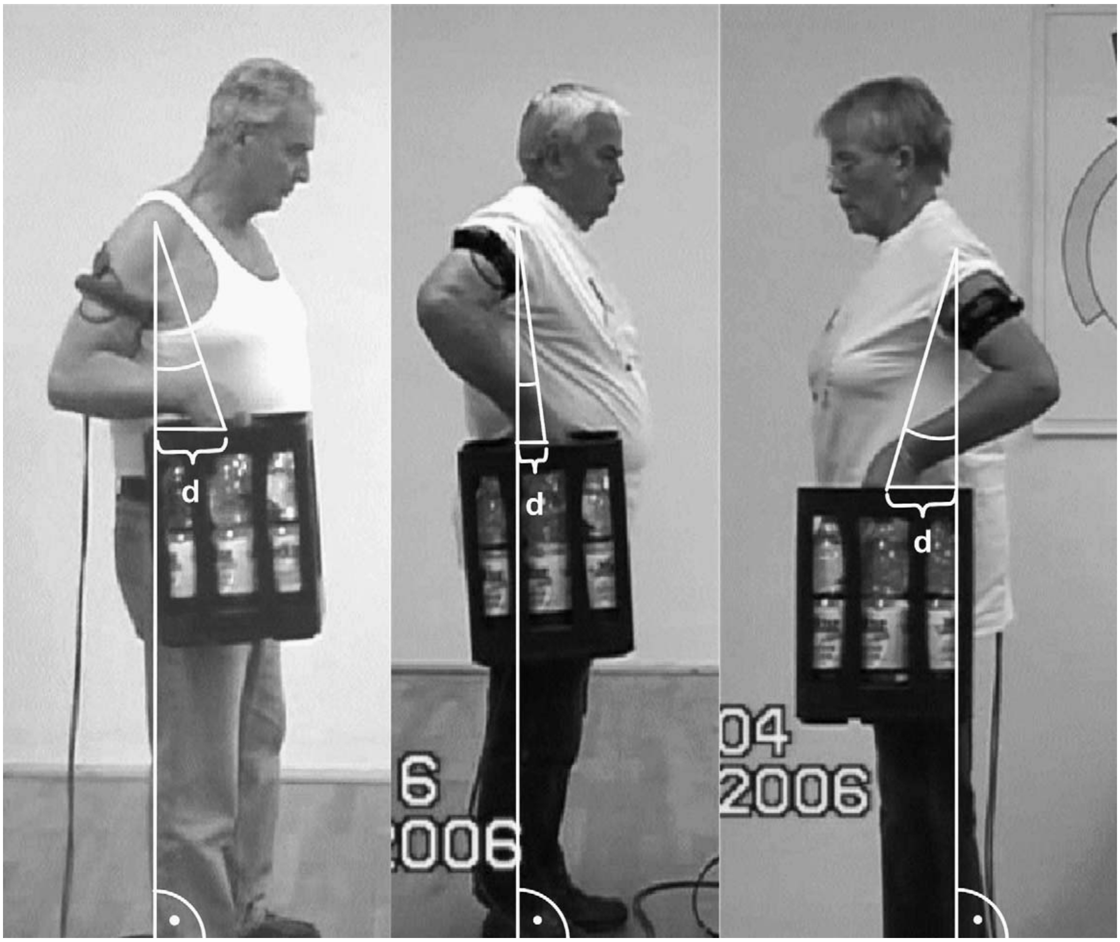

Fig. 5. Three patients during their lifting of 10 kg (Activity 5). Highest weight position for patients *S1R*, *S2R*, and *S3L*. *S2R* (middle) rotates his elbow sidewards; the weight has a short horizontal lever arm *d* relative to the shoulder joint center. *S1R* and *S3L* lift the weight more ventrally, thus creating a longer lever arm *d*.

3. Results

All average peak forces Fp and moments Mp are listed in Table 2.

1. *Combing* (Fig. 1): the manner in which combing should be carried out was not exactly specified, but all patients had a high elevation and external rotation of the arm during this task. Although no external loads acted during combing, Fp was always above 50%BW. It varied widely between 59.3% and 105.1%BW. This is a range of 60% relative to the value of 76.2%BW for the average patient *AP*. Peak moments Mp between 0.36% and 0.38%BWm were very similar in *S1R*, *S2R*, and *S4R*, while 0.64%BWm was measured in *S3L*.
2. *Two-handed steering* (Fig. 2): for steering with both hands, large inter-individual variations of Fp between 40.5% and 89.2%BW were observed. This is a range of 66% relative to Fp = 73.8%BW in *AP*. The highest moments and forces acted when the hand on the affected side was moved upwards (phases 2 and 3). The moment Mp for the male patients *S1R* and *S2R* remained low with 0.12% and 0.15%BWm, while the female patients *S3L* and *S4R* reached higher values of 0.53% and 0.36%BWm.
3. *One-handed steering:* the individual values of Fp were more consistent than during steering with both hands. Fp reached 106.1%, 137.3% and 123.9%BW for patients *S1R*, *S2R* and *S3L*. This is a range of 25% relative to Fp = 122.4%BW for *AP*, i.e. the forces vary much less inter-individually than for steering with both hands. The peak moment Mp of 0.40%BWm in *AP* was in the range observed for most other activities, and again *S3L* had the highest peak moment with 0.52%BWm compared to *S1R* (0.33%BWm) and *S2R* (0.35%BWm).
4. *Nailing* (Fig. 3): during nailing, the patients reached average peak forces between 80.3% and 117.1%BW. This is a range of 37.7% relative to Fp = 97.7%BW for *AP*. The contact forces caused by just lifting the hammer were higher than the additional forces required for hammering itself. The moments Mp reached 0.22%BWm for both male patients *S1R* and *S2R*, 0.54%BWm for *S3L* and 0.18%BWm for *S4R*.
5. *Holding and lifting* 10 kg *laterally* (Figs. 4 and 5): very low forces were found when the 10 kg crate was held passively (Fig. 4). The individual values of the resultant forces Fp varied between 9.2% and 17.9%BW. The force increased to peak values between 87.0% and 95.0%BW when the patients *actively* lifted the weight. This is a range of 8.7% relative to Fp = 91.5%BW for *AP*. The lifting heights were 38, 24, and 28 cm for *S1R*, *S2R*, and *S3L* (Fig. 5). In *S1R* and *S3L*, a maximum resultant moment of 0.46% and 0.62%BWm was measured, but in *S2R*, only 0.09%BWm acted.

   The patients *S1R* and *S2R* carried out this task also at a higher speed. In *S2R* the peak force Fp increased from 92.5% to 110.8%BW and Mp from 0.09% to 0.13%BWm. But the average lifting height was also higher with 31 cm compared to only 24 cm at slow speed. In contrast, the loads at higher speed decreased in *S1R* (Fig. 4). The peak force Fp was reduced from

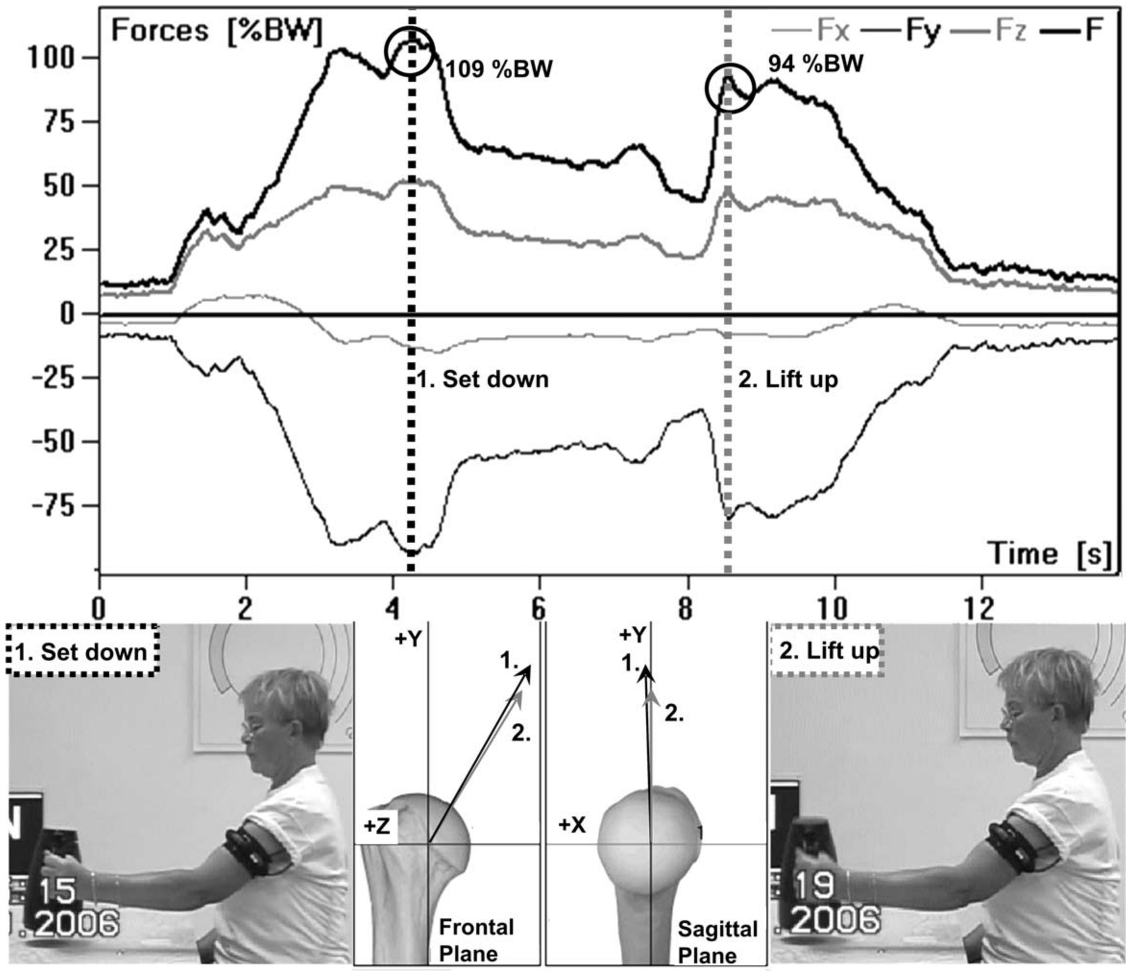

Fig. 6. Lifting up and setting down a coffee pot (Patient *S3L*; Activity 6). *Top*: force components and resultant force vs. time. *Bottom middle*: force directions in the frontal and sagittal plane. Patient lifts the pot close to the body (1–2 s), stretches out the arm (3.5 s), sets down the pot (5 s), remains in this posture (up to 8 s), lifts the pot (8.8 s), moves it back (up to 11 s), and sets it down (11.5 s). Patient's images and force directions (1. and 2.) in frontal and sagittal plane correspond to vertical dotted lines in diagram.

95% to 74.1%BW and the peak moment Mp from 0.46% to 0.37%BWm, while the average lifting height stayed nearly the same with 39 cm compared to 38 cm. The force patterns changed from a single peak shape to a double peak pattern (Fig. 4) for both patients and the maximum lifting height was reached exactly between the two peaks.

6. *Lifting the coffee pot* (Fig. 6): patients *S1R*, *S2R* and *S3L* reached forces Fp of 100%, 124.6% and 90.5%BW when lifting the pot with a straight arm. Relative to 105%BW for *AP*, this is a variation of 32%. Patient *S4R* was not able to perform this task due to muscle force deficiencies. Fp was higher for each patient when setting down the weight in the same position, and consequently also for *AP* with a value of 122.9%BW, although the mechanical situation was the same. The direction of Fp in the frontal plane was nearly identical for setting down as it was for lifting (Fig. 6, bottom). The moments Mp were nearly the same in all patients for lifting up and setting down. Patient *S3L* showed by far the highest resultant moments with 0.52% when setting down and 0.60%BWm when lifting up the coffee pot.
7. *Board at belt height* (Fig. 7): the forces Fp when setting down 2 kg differed to an extreme degree between 66.9% and 136.9%BW. This is a range of 80% relative to 88%BW for AP. The peak forces were already reached shortly before the weight touched the board. As for the coffee pot exercise, the forces during lifting the weight where always smaller than those during setting it down. The peak forces during lifting were reached exactly at the moment when the weight lost contact with the board.
The force directions throughout the whole motion (Fig. 7, top right) varied only slightly in the frontal plane (20°) and sagittal plane (22°), which was also seen at setting down and lifting the coffee pot. Even more consistent were the directions of the peak forces when lifting up and setting down the weight. They differed by only 5° in the frontal plane and 6° in the sagittal plane.
8. *Board at head height:* peak forces and moments both for setting the weight down and for picking it up increased between the task in belt height and in head height. During setting it down, for example, Fp rose from 94% to 118.8%BW in *S2R* and from 66.9% to 144.1%BW in *S3L*. As for the other exercises, lifting up the weight always caused smaller forces than laying it down. 15% less (103.6%BW) were measured in *S2R* and 51% less (93%BW) in *S3L*. The resultant moments were 0.25%BW for *S2R* and 0.64%BWm for *S3L* during putting in, and 0.33%BWm and 0.48%BWm for taking out the weight, respectively.

3.1. General findings

In general, the force–time courses were reproducible in one and the same patient, and quite similar even between different

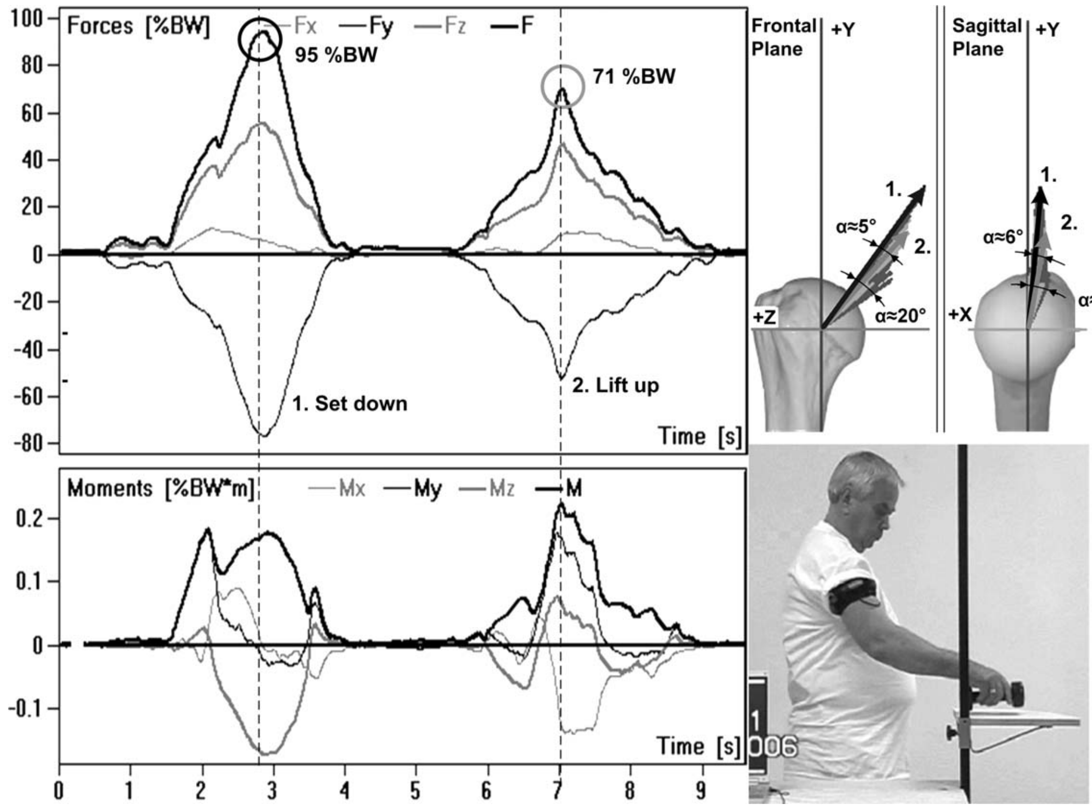

Fig. 7. Setting down and lifting 2 kg at belt height (Patient *S2R*; Activity 7). *Top left*: force components and resultant force vs. time. *Bottom left*: moments vs. time. *Top right*: force directions in three planes during the whole movement. Vectors (1. and 2.) correspond to the points of time (1. and 2.) of peak forces. *Bottom right*: patient's image at the instant of peak force (2.8 s). The arm is hanging down (0–1.4 s), the weight is picked up and lifted (2.2 s), the hand is moved forward and the weight positioned (2.8 s), the weight touches the board (2.9 s), the arm is moved back (3.8 s), and hangs down (up to 5.5 s). Then the arm is moved forward again (6.7 s), the weight is lifted (7.0 s), the arm is moved back (8.1 s), and the weight is laid down on the table (8.9 s).

Table 2
Forces and moments.

Activity		*S1R* (101 kg BW)		*S2R* (85 kg BW)		*S3L* (72 kg BW)		*S4R* (50 kg BW)		Average patient *AP*	
		Fp (%BW)	Mp (%BWm)	Fp (%BWm)	Mp (%BWm)	Fp (%BW)	Mp (%BWm)	Fp (%BW)	Mp (%BWm)	Fp (%BW)	Mp (%BWm)
1. Combing	Average (Std. dev.)	71 (4.4)	0.38 (0.09)	59.3 (2.2)	0.37 (0.01)	69.5 (4.6)	0.64 (0.11)	105.1 (1.04)	0.36 (0.01)	76.2 (3.1)	0.44 (0.06)
2. Steering with 2 hands		40.5 (3.77)	0.12 (0.01)	87.3 (16.9)	0.15 (0.05)	78.1 (2.68)	0.53 (0.02)	89.2 (4.48)	0.36 (0.05)	73.8 (7.0)	0.29 (0.03)
3. Steering with 1 hand		106.1 (8.14)	0.33 (0.03)	137.3 (11.95)	0.35 (0.08)	123.9 (6.76)	0.52 (0.03)	–	–	122.4 (9.0)	0.40 (0.05)
4. Nailing		80.3 (9.97)	0.22 (0.05)	117.1 (15.62)	0.22 (0.07)	83.5 (8.87)	0.54 (0.06)	110 (7.06)	0.18 (0.04)	97.7 (10.4)	0.29 (0.05)
5. Holding and lifting 10 kg	Holding weight	9.7 (4.8)	0.05 (0.02)	9.2 (3.86)	0.02 (0.01)	17.9 (4.94)	0.12 (0.01)	–	–	12.3 (3.4)	0.06 (0.01)
	Lifting slow	95 (25.48)	0.46 (0.09)	92.5 (9.64)	0.09 (0.02)	87 (10.4)	0.62 (0.09)	–	–	91.5 (11.4)	0.39 (0.07)
	Lifting fast	74.12 (14.98)	0.37 (0.12)	110.77 (9.34)	0.13 (0.03)	–	–	–	–	–	–
6. Coffee pot	Lifting	100 (2.18)	0.33 (0.04)	124.6 (6.64)	0.44 (0.07)	90.5 (7.86)	0.60 (0.02)	–	–	105.0 (5.6)	0.46 (0.04)
	Setting down	109.9 (3.12)	0.32 (0.02)	153.4 (6.0)	0.43 (0.07)	105.3 (4.94)	0.52 (0.04)	–	––	122.9 (4.7)	0.42 (0.04)
7. Board at belt height	Setting down	54.3 (11.67)	0.10 (0.02)	94 (3.19)	0.18 (0.01)	66.9 (5.03)	0.31 (0.03)	136.8 (7.44)	0.51 (0.03)	88.0 (6.8)	0.28 (0.02)
	Lifting	–	–	73.2 (6.56)	0.24 (0.01)	54.3 (0.79)	0.41 (0.03)	132.7 (0.29)	0.40 (0.01)	86.7 (1.9)	0.35 (0.02)
8. Board, head height	Setting down	–	–	118.8 (1.85)	0.25 (0.01)	144.1 (2.41)	0.64 (0.11)	–	–	131.5 (1.1)	0.45 (0.06)
	Lifting	–	–	103.6 (7.08)	0.33 (0.03)	93 (3.14)	0.48 (0.03)	–	–	98.3 (5.1)	0.41 (0.03)

Average values of peak resultant forces Fp and peak resultant moments Mp. Standard deviations are given in parentheses.

patients (Fig. 8, left side). In contrast to this, the moments varied more in the same subject and were often totally different among the individuals (Fig. 8, right side). It was a general observation from all activities that the inter-individual differences were much larger for the moments than they were for the forces.

4. Discussion

This study shows that the glenohumeral joint can frequently be loaded with more than one's own body weight during various activities of daily living. As a very simplified guideline one could

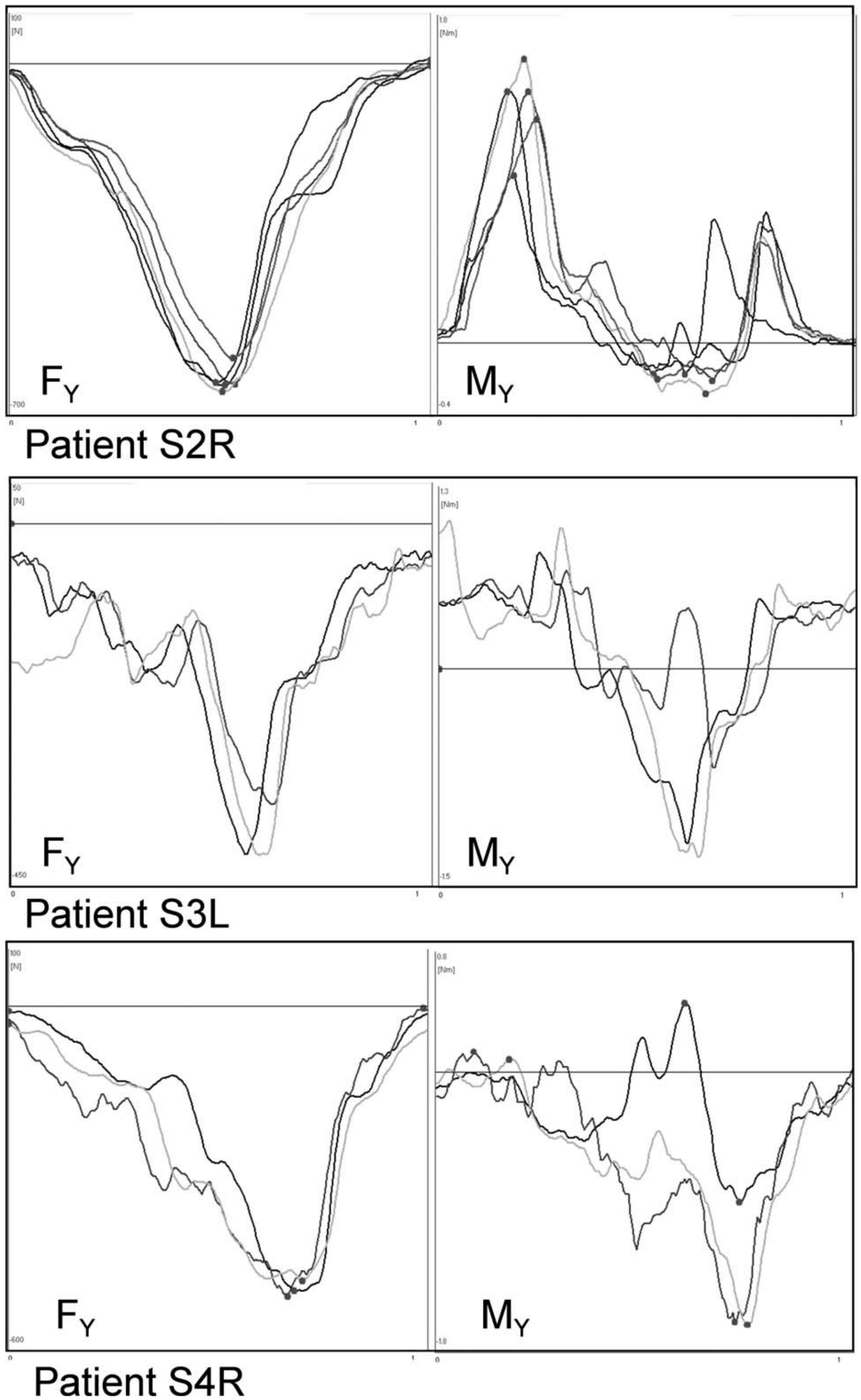

Fig. 8. Qualitative comparison of loads in three patients. Forces and moments in direction *Y* for *S2R*, *S3L* and *S4R* during the setting down of 2 kg on a board in belt height, 3–5 repetitions. The forces (left) show similar shapes inter and intra-individually. The moments (right) vary strongly between patients and even between trials.

say that the highest load levels to be seen can reach one third of those at the hip or knee joint.

This study is limited by the small number of elderly patients and because the trials could only be repeated 3–5 times in order to not overstrain the subjects. This prevents statistical evaluations of the data, but intra-individual comparisons of load magnitudes from different activities can still be performed. Comparisons between patients are very difficult due to wide varying age, weight and anthropometric data. Therefore the drawn conclusions between patients stay assumptions with minor statistical validity. The observed loads can only partially be transferred to young and healthy subjects. Despite these limitations, the obtained results are the only way to validate and improve musculoskeletal models. Based on our data, advice can now be given to patients and physiotherapists, as to which activities should be avoided shortly after implantation of a shoulder endoprosthesis or after fractures in this area. These activities would include lifting and setting down a weight with stretched arm or steering a car with one hand for example. The obtained results will also be applied to improve shoulder implants and fracture fixations in the shoulder region.

Compared to previous results (Bergmann et al., 2008), data from the same patients at later post-operative times has now been added, but only small variations were seen. This suggests that, as soon as a patient is able to fulfil a task, the load will not change considerably in relation to the post-operative time.

The variation of force directions in the course of most exercises was very small, especially for those activities which caused high peak forces: *nailing* (4), *coffee pot* (6) and *board* (7, 8). These only slightly varying directions relative to the humerus indicate widely changing directions relative to the glenoid or adaptive motions of the scapula. Similar small variations of peak forces directions relative to the femur have also been observed at the hip joint (Bergmann et al., 2001).

As suggested by Bergmann et al. (2007), the highest forces were observed whenever high external loads act at long lever arms (*board and coffee pot task, steering with one hand*) or when the active range of motion is reached, for example during the *combing task* (activity 1). De Wilde et al. (2004) showed that implantation of a prosthesis changes the humeral–scapular rhythm towards more motion in the scapula and less in the humerus. Compared to the optimized rhythm of healthy subjects, every change will therefore likely result in a less effective mechanism and therefore in higher contact forces like it was also seen at the hip (Bergmann et al., 1993).

During *combing* a high elevation angle and external rotation of the upper arm are necessary (Fig. 1; bottom right). Both movement directions are problematic for shoulder patients, so individual compensation mechanisms in the shoulder girdle probably lead to forces varying between patients.

During *steering* (2, 3) the highest forces occurred during upward movements of the arm. The force directions in the frontal plane (insert in Fig. 2) varied more throughout this exercise than they did in other tasks. From the mechanical point of view, force components transverse to the long axis of the humerus, as seen here, can cause high bending stresses in the proximal humerus which is critical for fresh fractures in this area, and can have adverse effects on the fixation of glenoid components. However, the force acts more in the humeral axis direction when its magnitude increases. The higher variation of peak loads when steering with both hands, compared to using one hand, can reasonably be explained by the possibility of load sharing between both sides.

The loads when passively *holding* 10 kg *beside the body* (5) were always less than 20%BW and even the values for lifting were much smaller than the 240%BW calculated by Anglin et al. (2000). Probably not only the muscle activities but also the scapula positions are adjusted to compensate the external weight and keep the glenohumeral joint nearly free of loads.

The contact force increased in all patients to uniform values of 87–95%BW when the external weight was *lifted* (5). However, the resultant moment remained very low in patient *S2R* with 0.09%BWm but went up to 0.46% and 0.62%BWm in the other two patients. Influencing factors could be individual friction properties of the synovial fluid, different glenoid shapes, or the horizontal lever arm of the weight, which was smaller in patient *S2R* (Fig. 5).

One possible explanation for the reduced forces observed in *S1R* when *lifting* the weight at a *higher speed* could be the vertical acceleration during the different phases of the movement. If a patient has sufficient muscular strength, he could accelerate the weight strongly in the beginning, causing the first peak (instant 3; Fig. 4). Then the weight moves further up to the highest point, while the contact force decreases (force minimum, instant 4). From there it is actively accelerated again downwards causing the second peak. Surprisingly there is no pronounced peak when the patient stops the weight at the end position. Compared to *S1R*, who felt the fast motion easier to perform, *S2R* was much more competitive and tried to reach the highest speed possible. This could explain the higher forces when he did the fast motion. Even for slow motion, if the weight was not held in the highest position, the force already decreases when the weight reaches the highest point (Fig. 4; instant 1+2).

During the *coffee pot task* (6) and the two *weight-into-board activities* (7+8) the forces were always higher when placing the weight down than when lifting it. This is probably caused by additional muscle activities, required to stabilize the hand position and to avoid a hard impact of the weight. Such stabilization effects are difficult to analyze using common, energy-related optimization criteria (Veeger et al., 2006); (Favre et al., 2005). Similar problems with antagonistic muscle activities also occurred with analytical models for other joints like the hip (Heller et al., 2001; Stansfield et al., 2003). Strategies other than using energy-related optimization criteria must probably be applied for quasi-static arm positions.

Another interesting observation for shoulder modelling is that patient *S3L* showed the highest moments during all tasks. One reason could be that she performed all tasks with her non-dominant arm, i.e. with insufficient muscular coordination. Moments generally vary more than forces because they are influenced both by variations of forces and variations of lever arms. If, in contrast to an idealized ball and socket joint without friction, the resultant contact force does not exactly pass through the center of the humeral head, this would change the lever arms. A contact force of 1000 N, for example, acting with an offset of only 4 mm would already cause a moment of 4 Nm; this is the range of maximum moments observed in our patients. An additional explanation for widely varying *joint moments* might be the individual-varying lubrication properties of the synovial fluid and the glenoid.

Conflict of interest

The authors declare that neither the authors nor members of their families have a current financial arrangement or affiliation with the commercial companies whose products may be mentioned in this manuscript.

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