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Elbow joint loads during simulated activities of daily living: implications for formulating recommendations after total elbow arthroplasty

Roos G.A. Duijn, Msc[a,*], Daniëlle Meijering, MD[b], Riemer J.K. Vegter, PhD[a], Friederike Albers, Msc[a], Alexander L. Boerboom, MD[b], Denise Eygendaal, MD, PhD[c], Michel P.J. van den Bekerom, MD, PhD[d,e], Martin Stevens, PhD[b], Reslin Schelhaas, Msc[a], Claudine J.C. Lamoth, PhD[a], Alessio Murgia, PhD[a]

[a]*Department of Human Movement Sciences, University Medical Center Groningen, University of Groningen, Groningen, the Netherlands*
[b]*Department of Orthopedic Surgery, University Medical Center Groningen, University of Groningen, Groningen, the Netherlands*
[c]*Department of Orthopaedics and Sports Medicine, Erasmus University Medical Center, Rotterdam, the Netherlands*
[d]*Department of Orthopedic Surgery, OLVG Hospital, Amsterdam, the Netherlands*
[e]*Amsterdam Movement Sciences, Faculty of Behavioral and Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam, the Netherlands*

Background: Overloading of the elbow joint prosthesis following total elbow arthroplasty can lead to implant failure. Joint moments during daily activities are not well contextualized for a prosthesis's failure limits, and the effect of the current postoperative instruction on elbow joint loading is unclear. This study investigates the difference in elbow joint moments between simulated daily tasks and between flexion-extension, pronation-supination, and varus-valgus movement directions. Additionally, the effect of the current postoperative instruction on elbow joint load is examined.
Methods: Nine healthy participants (age 45.8 ± 17 years, 3 males) performed 8 tasks; driving a car, opening a door, rising from a chair, lifting, sliding, combing hair, drinking, emptying cup, without and with the instruction "not lifting more than 1 kg." Upper limb kinematics and hand contact forces were measured. Elbow joint angles and net moments were analyzed using inverse dynamic analysis, where the net moments are estimated from movement data and external forces.
Results: Peak elbow joint moments differed significantly between tasks ($P < .01$) and movement directions ($P < .01$). The most and least demanding tasks were, rising from a chair (13.4 Nm extension, 5.0 Nm supination, and 15.2 Nm valgus) and sliding (4.3 Nm flexion, 1.7 Nm supination, and 2.6 Nm varus). Net moments were significantly reduced after instruction only in the chair task ($P < .01$).
Conclusion: This study analyzed elbow joint moments in different directions during daily tasks. The outcomes question whether postoperative instruction can lead to decreasing elbow loads. Future research might focus on reducing elbow loads in the flexion-extension and varus-valgus directions.

The study was reviewed and approved by the Medical Ethical Committee of University Medical Center Groningen, the Netherlands (METc2019/624).

*Reprint requests: Roos G.A. Duijn, Msc, Department of Human Movement Sciences, University of Groningen, University Medical Center Groningen, Antonius Deusinglaan 1, Huispostcode FA23, Groningen 9713 AV, the Netherlands.

E-mail address: r.g.a.duijn@umcg.nl (R.G.A. Duijn).

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Total elbow arthroplasty (TEA) is a surgical procedure that is, performed to reduce pain and regain function in patients with a variety of debilitating elbow pathologies, such as inflammatory or post-traumatic arthritis and complex fractures.[38] Although the use of TEA is growing, it remains a relatively uncommon orthopedic procedure, with 3146 TEAs performed over a 5-year period in the USA.[41] In the Netherlands, the number of TEAs rose from 67 in 2017 to 73 in 2022.[19] It is performed more often in women than in men.[7,13] Unfortunately, elbow prosthesis survival rates following TEA are low compared with those following hip and knee arthroplasties. The current TEA survival rate in the range of 7.5-14.2 years is 71%-87%[31,37,38] compared to a 10-year survival rate of 90%-95% for hip and knee arthroplasty.[8,17,33]

The most important factor determining poor survival rates of elbow arthroplasty is aseptic loosening of the humeral component.[31,32,36-38] Retrieval studies showed that aseptic loosening results from different mechanisms. First, overloading of the prosthesis during activities of daily living (ADL) causes polyethylene (PE) wear,[26,32] bringing loose particles to the proximity of the bone-prosthesis interface, ultimately leading to bone destruction and inflammation.[4,12,32] Revision surgery becomes necessary as a result of aseptic loosening. A second mechanism of failure may be inadequate cement fixation,[22] where excessive stresses at the bone-cement-prosthesis interface lead to micromotion and consequent failure of the implant. Last, joint replacements face greater durability requirements as life span lengthens. It has been observed that patients following TEA stay active for a longer period and are less inclined to restrict their lifestyle to safeguard the prosthesis's lifetime.[2] In the long term, this behavior may lead to overloading of the implant during activities of daily living.[21]

In vitro tests to analyze elbow joint biomechanics have provided insight into the failure mechanisms of TEA. Lo and Lipman examined retrieved Coonrad-Morrey ulnar components, demonstrating that varus-valgus (VV) moments in the ulnohumeral joint as high as 5 Nm would lead to loads exceeding the PE's theoretical yield strength of the Coonrad-Morrey prosthesis and consequently to irreversible plastic deformation.[18] Furthermore, elbow joint loading of 3 different types of TEAs while holding a 2.3-kg weight exceeded the yield strength of PE when the shoulder was abducted at 45° and 90°.[15] However, these results are not easily generalizable to ADL tasks, as it is not known what the specific loads during different ADL tasks are and which tasks could lead to a potential overload of the elbow joint.

To reduce overloading of the elbow, common clinical practice is to instruct patients not to perform daily activities that regularly exceed lifting 1 kg weight and incidentally up to 5 kg.[5,9] However, this guideline lacks specificity because it is unclear which ADL tasks or specific movements would exceed the 1-5 kg limits and should thus be avoided.[5] Depending on the type of movement and how it is executed, similar weights can lead to different loads on the elbow.[15,20] It is therefore crucial to know whether the current clinical instruction brings about a detectable reduction in elbow joint loads.

The low survival rates, combined with the lack of consensus on postoperative management, emphasize the need for biomechanical studies focusing on elbow joint loading during ADL. The aims of the current study are as follows: first, to identify any differences in range of motion (ROM) and elbow joint moments between 8 ADL tasks in a lab setting; second, to identify differences in peak loads between the flexion-extension (FE), pronation-supination (PS), and varus-valgus (VV) movement; and last, examine whether the current instruction of "not lifting more than 1 kg" leads to a decrease in elbow joint moments. We hypothesize that joint moments will differ per task and that the instruction will not lead to a decrease in elbow joint load.

Materials and methods

Participants

Nine healthy, able-bodied participants performed 8 simulated ADL tasks. Exclusion criteria were (1) mental or physical disability to meet study requirements; (2) insufficient command of the Dutch language; and (3) prior surgery in the upper extremity or other pathologies affecting upper extremity function. Participants were informed about the procedures and signed an informed consent. Anthropometric data are presented in Table I.

Procedure

Basic anthropometric data (height, weight, arm length, shoulder width) and a maximum voluntary contraction of the biceps and triceps were collected at entrance. After a static calibration trial, participants performed a standardized series of 8 ADL tasks in 2 conditions (Fig. 1). All tasks are explained in Table II. After one entire series (uninstructed condition), another series followed with each task performed again (Fig. 1), this time with a verbal instruction comprising the recommendation to "not exceed lifting 1-kg weight and only incidentally use up to 5 kg" (instructed

Table I Participant demographics

	n	Mean	SD
Age (yr)	9	45.8	17.4
Height (cm)	9	178	7.4
Weight (kg)	9	74.9	5.4
Shoulder width (cm)*	9	44.4	2.1
Arm length (cm)†	9	65.9	2.8
Gender			
Male	3		
Female	6		
Dominance (%)‡			
Right	8	86.1	12.4
Left	1	−90	

SD, standard deviation.
* Measured from left acromion to right acromion.
† Measured from dominant acromion to third metacarpophalangeal joint of the dominant arm.
‡ Handedness was analyzed using the Edinburgh inventory (Oldfield et al, 1971).

condition). In both conditions, the tasks were performed in the same fixed order (Table II).

The tasks were selected based on the expected amount of elbow movement, as well as on patients' frequently asked questions after TEA surgery.[1] In the seated tasks, the participant sat on a height-adjustable chair without back support. A 75-cm-high table was placed in front of the participant, with a marked starting point. The height of the chair was adjusted so that the elbow was flexed at 90° when the hand was placed on the table, and the upper arm was held in vertical position (Fig. 2). For each task, an initial position and aim were defined. After verbal instruction and 1 test trial, each task was repeated 5 times. The participant was instructed to move at a comfortable speed throughout the experiment. Between the different tasks, there was a rest period of at least 30 seconds. All tasks were performed consecutively twice, ie, in 2 consecutive conditions.

In 3 ADL tasks (1, 2, 3), a force transducer was used to record generated external reaction forces. During the steering wheel task, a constant force of 15 N for the car task and 25 N for the door task was applied as resistance.[16,25]

Instruments and data collection

Body segment position of the upper extremity was collected at 100 Hz using a 4-position sensor motion capture system (Optotrak 320; Northern Digital Inc., Waterloo, ON, Canada). Four infrared light-emitting markers were placed on bony landmarks of the upper limb and thorax. Six rigid bodies were placed on the thorax and upper limb segments, which mapped 14 additional virtual markers. Last, 1 marker was placed on the center of the force transducer and 1 marker on the 1-kg object. All marker positions are shown in Appendix A. The coordinate system of the marker data was set on the table, with X forward, Z to the right, and Y upward.

Force data were recorded with a force transducer (ME-Messysteme GmbH, Henningsdorf, Germany) with an accuracy of 0.01 N. The force transducer was mounted on an aluminum T-bar and could be set in different positions (Fig. 2, *A* and *B*). Both marker data and force data were recorded with a frequency of 100 Hz. The motion capture system and the force transducer were digitally synced to enable simultaneous recording.

Data analysis

The force values in the local coordinate systems were converted into a global coordinate system of the motion capture system using a customized MATLAB script (version 20a; MathWorks Inc., Natick, MA, USA). The motion capture data and force data were filtered in MATLAB using a fourth-order Butterworth filter with a 6-Hz cutoff frequency. Data gaps were reconstructed using piecewise cubic spline interpolation.

Musculoskeletal model

OpenSim musculoskeletal modeling software (version 3.3; Stanford University, Stanford, CA, USA) was used to run the dynamic Holzbauer model.[11,35] This model consists of 7 bone segments and 50 Hill-type muscle-tendon actuators, representing 32 muscles and muscle compartments (Fig. 2, *C*). This model, which only allowed elbow joint moments in the FE and PS direction calculation, was adjusted to include the VV moments. To analyze the VV direction, an extra degree of freedom was computed in the humeroulnar joint. The maximum VV ROM was set from −11.2° valgus to 6.6° varus.[29] The model was scaled in OpenSim to the body dimensions of the participant using data from the static calibration trial. The anatomic locations and the segments' coordinate systems are in accordance with the International Society of Biomechanics recommendations.[40] Inverse kinematic analysis was accomplished by using the OpenSim application programming interface in MATLAB.[6] Inverse dynamic analysis was performed in OpenSim software interface. The external reaction forces were applied to the model's hand segment at the distal end of the third metacarpal bone of the dominant hand. The gravitational force was applied to the hand on the task where the 1-kg object was lifted. In the task where a 1-kg object was slid across the table, a dynamic friction coefficient of 0.5 for polyethylene-on-polyethylene was applied. Repetition 2, 3, or 4 was normalized over time from the start of the movement to the end.

The data were further processed in MATLAB 20a to extract (1) the ROM in the FE and PS directions and (2) peak elbow joint moments in 3 different directions (FE, PS, VV). The standard deviations of the peak elbow joint moments were calculated to quantify intersubject variability.

Statistical analysis

Peak joint moments were used to test the differences in elbow joint moments between direction (FE, VV, PS), tasks (car, door, and chair), and condition (instructed, uninstructed). To analyze the data, IBM SPSS Statistics, version 29 (IBM Corp., Armonk, NY, USA), was used for linear mixed-model analysis. Bonferroni correction was used for post hoc tests. Cohen *d* was calculated to indicate the effect size of the post hoc tests. A P value $< .01$ was considered statically significant. Normality of data distribution was tested to allow parametric testing.

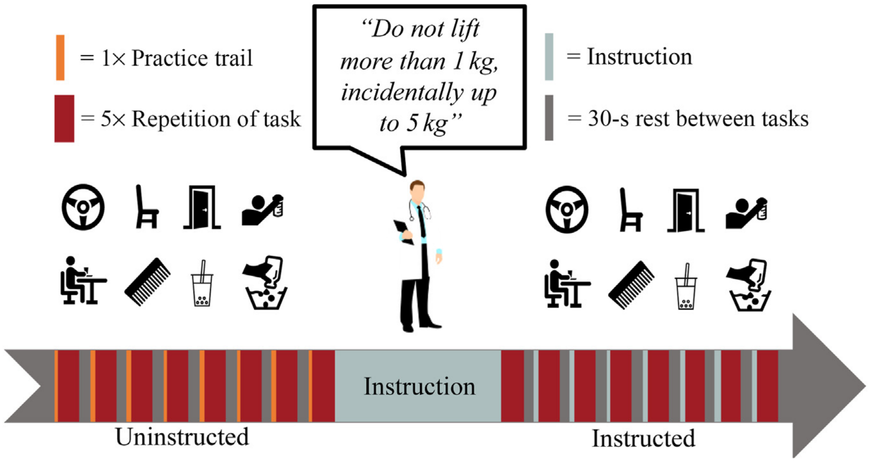

Figure 1 Research protocol. Each symbol illustrates an ADL task. Participants first performed a series of 8 ADL tasks. Next, there was the instruction of "not lifting more than 1 kg." Then the second series was performed. The instruction was repeated before each task. *ADL*, activity of daily living.

Because the joint moments were not normally distributed, the positively skewed distribution of joint moments was normalized via a square root transformation (Appendix B) so that the linear mixed model could be used.

Results

Kinematic data

The average ROM of all the participants is shown in Fig. 3. Chair, door, and cup tasks showed the largest FE-ROM (range 19°-110°). The hair and drinking task corresponded with the highest extension angles (126°). Large PS-ROM was observed during the cup task (range 38° to −59°). The greatest ROM variability was seen during the hair task.

Joint moments

Fig. 4 shows the normalized joint moments over time in the FE, PS, and VV directions for the 8 ADL tasks. In particular, the chair, car, and door tasks show greater intersubject variability. Peak elbow joint moments for the FE, PS, and VV direction are shown in Table III. The overall highest peak moments were observed when rising from a chair (13.4 Nm extension, 5.0 Nm supination, and 15.2 Nm varus), followed by steering a car (9.3 Nm extension, 5.4 Nm supination, and 15.2 Nm valgus). The slide task required the smallest elbow moment (4.3 Nm flexion, 1.7 Nm supination, and 2.6 Nm valgus). Table IV shows an overall ranking of the 8 tasks based on the joint moment magnitude and the externally applied force. Greater elbow joint moments, in all directions, were present during those tasks with an external reaction force applied on the hand.

Statistical outcomes are presented in Table V. Peak elbow joint moments were significantly different between the tasks, $F(7, 376) = 40.44$, $P < .001$. There was a significant difference in elbow joint loads between the movement directions, $F(2, 376) = 170.02$, $P < .001$. Post hoc test showed that VV moments ($P < .001$, Cohen $d = 1.3$) and FE moments ($P < .001$, Cohen $d = 1.3$) were significantly higher than PS moments.

The instruction did not lead to a significant decrease in elbow joint load, $F(1, 376) = 2.07$ $P = .15$. However, a significant interaction effect of task and condition was found ($P < .01$). This evidences that the instruction only had an effect on selected tasks. Follow-up analysis showed that only during the chair task was there a significant decrease in elbow joint load when the instruction was followed, $t(1, 376) = 2.58$, $P < .01$. During this task, the participants lifted their own body weight using a combination of arm and, possibly, leg movements.

Discussion

In this study, 8 simulated ADL tasks were analyzed on ROM and peak joint moments in the FE, PS, and VV direction, depending on the given verbal instruction of not lifting more than 1 kg. Joint moments differed between tasks and movement directions. FE moments and VV moments were significantly higher compared with PS moments. The effect of the instruction of "not lifting more than 1 kg" was dependent on the tasks. Only during the chair rise task did the instruction result in a significant decrease in elbow joint loads. These results confirm our hypothesis that elbow joint moments differ per task and indicate that the current instruction might be reconsidered to emphasize the load demand per task based on biomechanical evidence.

Table II Description of activities of daily living and order of execution

	Activity	Initial position	Aim
	1. Steering a car wheel (T-bar horizontal on the table).	Table placed in front of participant, at 1 AL. Chair is lowered, legs stretched. Dominant hand on T-bar, nondominant hand on the leg.	Turn T-bar, using the handle, from 10 o'clock to 4 o'clock and turn it back to 10 o'clock.
	2. Opening and closing a door (T-bar vertical on the table).	Participant stands in front of the door, with elbow at 90°, hand resting on the handle.	Push T-bar to 90° using the handle. Close the door, back to starting position.
	3. Rising from a chair (T-bar in armrest).	Seated, both hands on armrests.	Rise from the chair using armrests, sit down again.
	4. Lifting 1-kg object	Target X on the platform (SH) placed 1 AL and 1 SW from dominant arm. Hold 1-kg object at SP.	SP, place object at target X, back to SP.
	5. Sliding 1-kg object	Target X at 1 SW on the nondominant side, 1 AL with the dominant arm. Hold 1-kg object at SP.	SP, slide target to target X, and back to SP.
	6. Combing hair	Rest a hand on SP.	SP, combing hair in the midline back and forth, SP.
	7. Drinking (1 kg)	Hold 1-kg object at SP.	SP, simulate drinking, SP.
	8. Emptying a cup (1 kg)	Target 1 full AL in front of participant. Hold 1-kg object at SP.	SP, stretch arm toward target, 180° rotation counterclockwise, then back, SP.

AL, arm length, measured from acromion to third metacarpophalangeal joint of the dominant hand; *SH*, shoulder height; *SW*, shoulder width, measured acromion-to-acromion; *SP*, starting point.
The T-bar is an aluminum bar connected to the force transducer. Each activity was repeated 5 times, all within 35 s.

The tasks performed during this study give a good representation of elbow ROM needed to naturally perform ADL. The findings of the FE-ROM (range 19°-126°) are in line with earlier literature.[27] A review by Oosterwijk et al[27] concluded that an FE-ROM 0°-150° is required for ADL tasks, which is more than the generally used reference of 30°-130°.[21,34] Mainly tasks needed for personal care and feeding needed a flexion angle >135°.[27] FE angles <30° were observed during the door, cup, and lift tasks. A review of Kincaid and An shows that especially peak bone-on-bone contact forces occur between 7° and 11° flexion (almost fully extended).[14] Muscle activity and therefore bone-on-bone contact forces would be higher early in the flexion cycle because of the poor mechanical advantage of the prime movers: the brachialis, biceps, and brachioradialis muscles.[24] The functional PS-ROM of −50° pronation to 88° supination found in our study is higher than the PS-ROM found by Sardelli, who reported a functional ROM between −65° ± 8° pronation and 77° ± 13° supination.[34]

In this study we found that overall peak FE moments and VV moments were higher compared to the peak PS moments. Therefore, the focus in preventing overload should probably be on reducing elbow moments, mainly in the FE

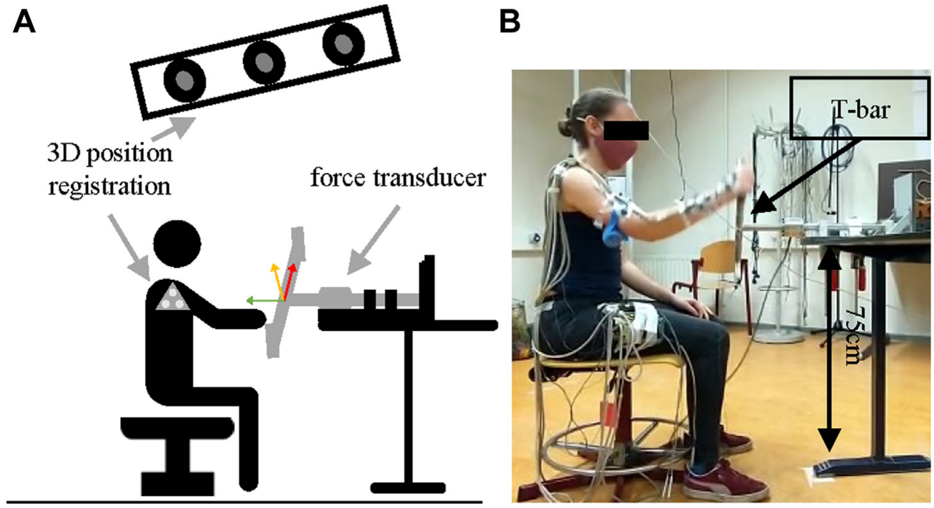

Figure 2 Schematic overview of experimental setup for car task with the force transducer (15-N resistance force). (**A**) Sagittal view of the lab setting. (**B**) View of the geometry of the musculoskeletal model. (**C**) View during the car task. The markers are in pink dots, the muscles in red, the ← represents the external reaction force (force from T-bar to hand).

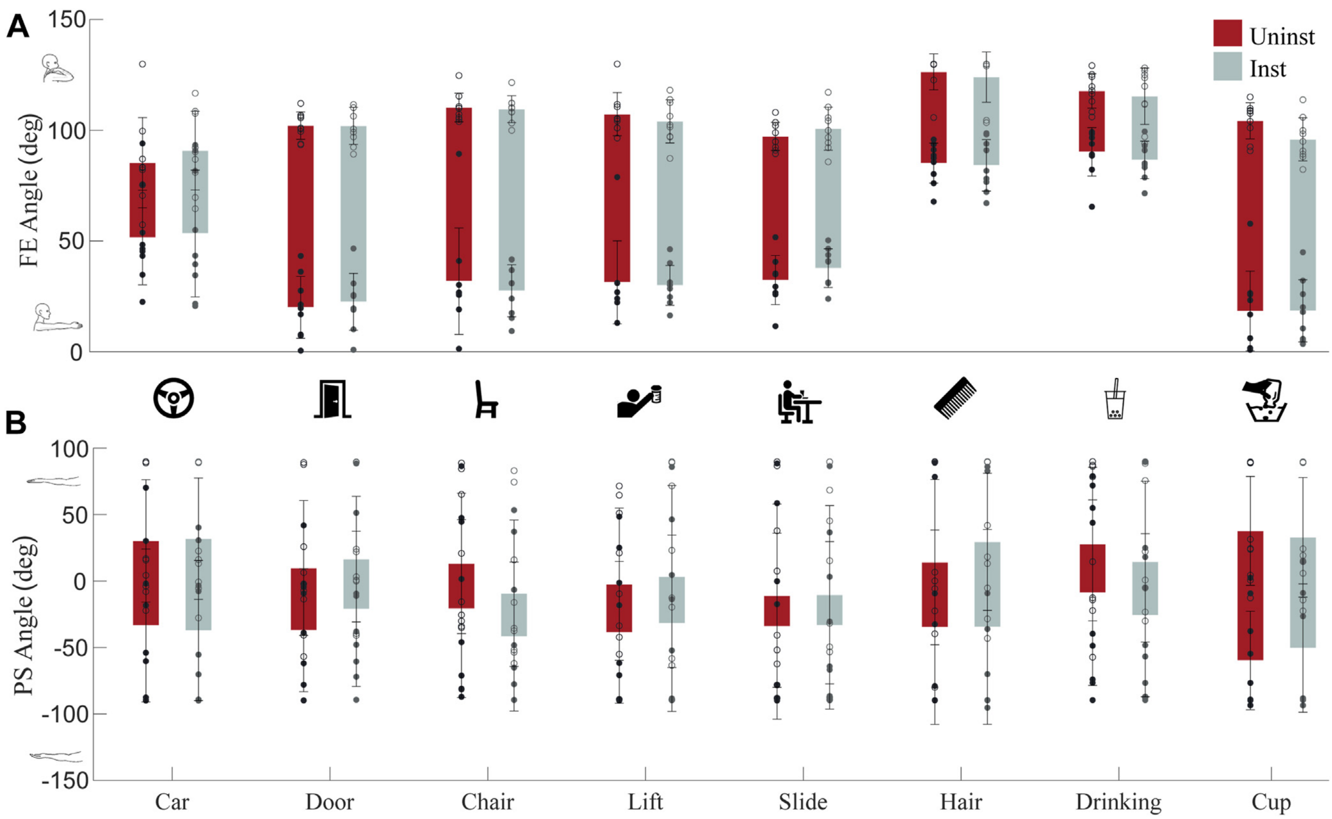

Figure 3 Average (n = 9) range of motion in (**A**) flexion-extension (0° is fully extended) and (**B**) pronation-supination (90° is fully pronated) direction for 8 simulated activities of daily living. Without instruction (*red*) and with instruction (*gray*) of not lifting more than 1 kg. Dots represent minimal individual angles (*filled*) and maximal angles (*empty*). Angles are in degrees. Error bars represent the standard error for the maximum angle (upper end) and the minimal angle (lower end). *FE*, flexion-extension; *PS*, pronation-supination.

and VV directions. The high VV-moments found, especially during rising from a chair and steering a car, are likely due to the combination of large external reaction forces and a relatively large moment arm.

Although no other study examined ADL tasks in all 3 directions, results of previous research on FE and PS moments are comparable to our findings. For instance, the task performed by Murray et al[23] showed the same order of

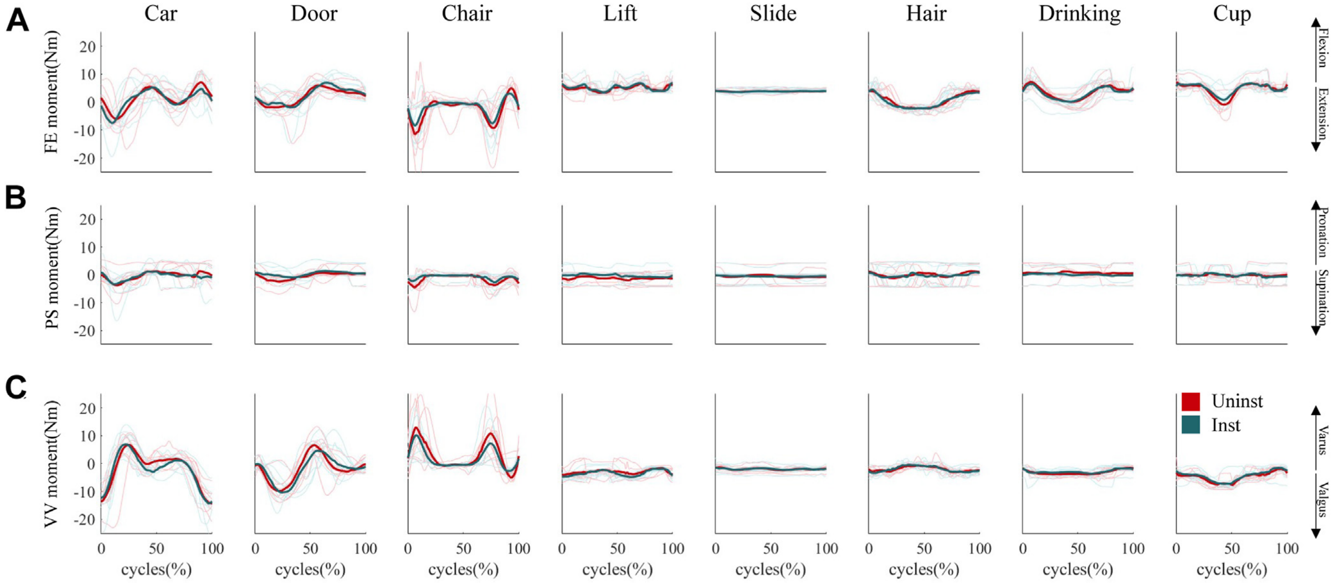

Figure 4 Average (n = 9) (*dark*) and individual (*light*) normalized elbow joint moment for the selected ADL tasks (1 of 5 repetitions). *Red*: uninstructed condition, *blue*: instructed condition. (**A**) Elbow joint FE moment. Negative values indicate extension moment, and positive values, flexion moment. (**B**) Elbow joint PS moment. Negative values indicate supination moment, and positive values, pronation moment. (**C**) Elbow joint VV moments. Positive values indicate varus, and negative values, valgus. One of 5 repetitions of every task was normalized over time. *FE*, flexion-extension; *PS*, pronation-supination; *VV*, varus-valgus; *ADL*, activities of daily living.

Table III Peak elbow joint moments (N = 9) in an uninstructed and instructed condition for 8 simulated activities of daily living

Task	FE moment, Nm*				PS moment, Nm†				VV moment, Nm‡			
	Extension (−)		Flexion (+)		Pronation (+)		Supination (−)		Varus (+)		Valgus (−)	
	Uninst.	Inst.	Uninst.	Inst.	Uninst.	Inst.	Uninst.	Inst.	Uninst.	Inst.	Uninst.	Inst.
1. Steering a car wheel	−8.6 (4)	−9.3 (4)	9.1 (2)	8.5 (2)	2.5 (2)	3.3 (3)	−4.8 (3)	−5.4 (4)	8.1 (3)	8.0 (3)	−14.2 (4)	−15.2 (6)
2. Opening and closing a door	−4.1 (4)	−3.7 (5)	7.2 (2)	8.1 (2)	1.8 (1)	2.1 (2)	−3.1 (2)	−1.8 (2)	7.7 (2)	7.8 (2)	−10.2 (2)	−11.0 (4)
3. Rising from chair	−13.4 (6)	−10.3 (7)	6.4 (3)	3.1 (2)	1.3 (1)	0.6 (1)	−5.0 (3)	−3.4 (2)	15.2 (10)	11.1 (6)	−5.6 (3)	−3.0 (1)
4. Lifting 1-kg object	—	—	7.5 (2)	7.7 (2)	2.7 (1)	1.1 (2)	−0.1 (2)	−1.6 (2)	—	—	−5.2 (1)	−5.3 (1)
5. Sliding 1-kg object	—	—	4.2 (1)	4.3 (1)	0.4 (2)	-	−1.7 (3)	−1.4 (2)	—	—	−2.6 (1)	−2.6 (1)
6. Combing hair	−3.2 (1)	−3.2 (1)	5.6 (1)	5.2 (3)	1.5 (2)	1.4 (3)	−2.0 (3)	−1.6 (3)	0.4 (1)	0.1 (1)	−4.2 (1)	−4.2 (2)
7. Drinking	−0.3 (2)	−0.6 (3)	8.4 (2)	7.7 (3)	1.7 (2)	0.9 (2)	−0.6 (2)	−0.9 (2)	—	—	−4.4 (1)	−4.3 (2)
8. Emptying cup	−1.3 (3)	—	8.0 (2)	7.7 (2)	1.8 (2)	1.9 (2)	−2.6 (2)	−2.6 (2)	—	—	−7.8 (1)	−7.7 (1)

FE, flexion-extension; *Uninst.*, uninstructed; *Inst.*, instructed; *PS*, pronation-supination; *VV*, varus-valgus.
Values within parentheses are standard deviations.
* Peak flexion and extension elbow joint moment in newton meters for the uninstructed an instructed tasks. A negative value indicates an extension moment, and a positive value a flexion moment.
† Peak pronation and supination elbow joint moment in newton meters for the instructed and uninstructed tasks. A positive value indicates a pronation moment, and negative value, a supination moment.
‡ Peak varus and valgus elbow joint moment in newton meters for the instructed and uninstructed tasks. A positive value indicates a varus moment, and negative value, a valgus moment.

magnitude as those of our study, although we used a 1-kg object compared to Murray's 0.5-kg object, resulting in higher FE moments. To illustrate, during the lifting task, Murray et al found a maximum FE moment of 5.8 Nm during the lifting of a block (∼0.5 kg) to head height, whereas in our study, where the object was placed at shoulder height, a flexion moment of 6.4 Nm was required. The results of Cheng et al[3] were comparable to ours,

Table IV Ranking for each task based on joint moments and external force

Task	FE moments*	PS moments*	VV moments*	External loads†	Overall total
1. Steering a car wheel	2	1	1	2	6
3. Rising from chair	1	2	2	1	6
2. Opening and closing a door	3	3	3	3	12
8. Emptying cup	5	4	4	4	17
7. Drinking	4	7	6	4	21
4. Lifting 1-kg object	6	6	5	4	21
6. Combing hair	7	5	7	8	27
5. Sliding 1-kg object	8	8	8	7	31

FE, flexion-extension; *PS*, pronation-supination; *VV*, varus-valgus.
Lower values indicate a higher risk of polyethylene wear.
* Elbow joint moments in FE, PS, and VV direction, ranked from high to low; higher joint moments indicate a higher risk of polyethylene wear.
† Total external reaction force, calculated with output force transducer or gravitational/friction force. Higher external load indicates a higher risk of polyethylene wear.

Table V Statistical outcomes of the linear mixed models

	df	*F*	*P* value
Tasks	7	40.44	<.001
Direction	2	170.02	<.001
Condition	1	2.07	.151
Direction × Condition	2	0.08	.992
Condition × Tasks	7	3.04	<.001
Task × Direction	14	6.29	<.001
Tasks × Direction × Condition	14	0.31	.993

df, degrees of freedom.
Joint moments were compared between different tasks, directions, and conditions. *P* value < .01 was considered as statically significant.

although they found higher moments in tasks where they used a 2-kg object compared to our results, where a 1-kg object was used.

For all participants, the highest peak moments, in FE direction, were achieved during the rising phase of the chair task. The lowest elbow joint moments were observed during the slide task. King et al[15] showed that the amount of shoulder abduction affects the joint loading of the elbow. The variation in elbow moment is surprisingly low in the slide task compared to the lifting and cup tasks (which show the same FE-ROM). This may indicate that all participants used the same movement strategy during the slide task, which is initiated from the shoulder. More research is needed to elucidate whether shoulder moments could partially relieve elbow joint loading during selected tasks.

Altogether, the chair, door, and car tasks showed the highest risk of wear because of the higher observed external reaction forces resulting from pushing and pulling. These findings question whether the focus of the instruction should be on lifting an object (“not lifting 1 kg”) or on the presence of external reaction force, ie, the amount of force required for a pulling or pushing movement.

Instruction

No overall main effect of instruction was found; however, the *P* value was low ($P = .15$). The results show that the effect of the instruction is dependent on the tasks. During the chair task, a significant change was found after the instruction in FE moments (13.4-10.3 Nm) and VV moments (15.2-11.1 Nm). This was the consequence of less pushing force against the armrest, possibly combined with a greater leg pushing force.

Contrary to our expectations, during the door and lift tasks unexpected higher peak flexion moments were found during the instructed condition, with a change in the minimal FE angle during the door task (Fig. 3, *A*). However, this kinematic change did not lead to a significant decrease in elbow joint moments. This finding questions whether people can accurately predict which changes in movement will lead to lower loads in the elbow.

Implications for implant failure

Finite element studies indicated that VV moments of 5 Nm at the ulnohumeral joint would possibly exceed the yield strength of PE. Surprisingly, the overall mean of the VV moments found in this study (7.9 Nm) exceeded these failure limits, leading to permanent deformation of the PE material. Moreover, during 5 (car, door, chair, lift, and cup) of the 8 tasks, even higher external VV moments were observed (Table III).

Besides the loading, task frequency also plays an essential role in the risk of PE wear. Many repetitions of a movement lead to erosion of the material. The frequency of the FE movement associated with normal ADL is estimated to be 0.5 million cycles/year, whereas for strenuous ADL with a significant weight in hand the frequency is approximately 7500 cycles/year.[30] It is therefore important to remember that there is not one specific limit or threshold on whether a task can be performed, as it is also important to consider the frequency of the task.

Recommendation for clinical practice

So far, it is known that high elbow loads lead to PE wear, which ultimately causes permanent deformation of the prosthetic PE material. Based on the results of the current study, we can now give a better indication of which tasks are more demanding. First, frequent repetition of heavy tasks with a large amount of external load should be avoided or performed differently than before the operation, eg, rising from a chair without using the armrest instead of pushing with whole-body weight. Hence, external loads are not only the loads resulting from lifting an object (ie, a heavy book or groceries) but also from the reaction force on the object. Plus, tasks further away from the body with an outstretched arm (ie, reaching) are more demanding than tasks closer to the body.[20]

Limitations and future directions

The musculoskeletal model used in the current study was based on a single cadaver specimen. Individual differences could have led to soft tissue artifacts or incorrectly defined joint centers in marker-scaled models, affecting the inverse dynamics results.[10] However, the elbow axes and kinematics were defined in accordance with the International Society of Biomechanics recommendations,[40] and therefore the effect of individual differences is subsequent to the correct anatomic behavior. Future research could further investigate the effect of individual morphologic differences on the inverse dynamics estimations.

Second, although healthy participants were examined in this study, it is possible that elbow motions following TEA are changed because of altered motion pathways (ie, rotation axis), proprioception, and muscle forces.[2] Future research should examine the changes in elbow joint moments in TEA patients and incorporate the changes in prosthesis kinematics into the musculoskeletal model. Computed tomography scans, combined with artificial intelligence technology, have already been used to measure muscle elongations for different implants, positions, and patient anatomies, and can therefore be used to personalize the model.[28]

Last, we do not know if the observed elbow loads of the current study lead to failure of the material of the elbow prosthesis. So far, we could only compare 1 feature of the elbow load (VV moments) to the reported failure, ie, permanent deformation and limits of the PE material of one specific elbow prosthesis. Besides, different types of prostheses may have different failure mechanisms and limits.[39] To compare the in vivo elbow load to the loads that exceed the failure limits of the prosthetic material, future research should focus on failure limits and elbow joint load in both elbow joint moments and internal bone-on-bone contact forces.[20]

Conclusion

Results of the current study provide insight into elbow joint loading during ADL tasks. Tasks that include pushing and pulling result in higher joint loads, especially in the FE and VV direction. Surprisingly, the VV moments found in this study exceeded the failure limits, leading, theoretically, to permanent deformation of the prosthetic material. The current study found that joint moments could provide a loading range for in vitro testing of prostheses during the design stage. To avoid overloading the elbow prosthesis, the current postoperative instruction does not appear to be sufficient. The outcomes of this study can be used as a first step in formulating evidence-based and specific instruction. However, bone-on-bone contact forces and elbow joint moments (VV, PS, and FE direction) in both healthy adults and patients following TEA need to be further analyzed to draw more definitive conclusions on elbow joint loading in ADL.

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Supplementary data

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