Kinematics and stability of the Norway elbow

A cadaveric study

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We investigated the effect of simulated muscle loading and the contribution of the radial head to stability of the Norway elbow in 6 cadavers using an electromagnetic tracking device. The kinematics of the elbow after implantation of the prosthesis were similar to the intact elbow in their valgus-varus orientation, however, the forearms were slightly externally rotated, probably due to a small amount of external rotation of the humeral components at the time of implantation. The valgus-varus laxity limit of the implants were greater than in the intact specimens averaging 8.0 and 5.6 degrees, respectively. Simulated muscle loading stabilized both the intact and the Norway elbows. Excision of the radial head after implant arthroplasty increased their valgus-varus laxity, suggesting that preservation of the radial head may be indicated if it is not too severely involved by the underlying disease process.

The laxity permitted by the prosthesis articulation is greater than that measured after implantation of the Norway arthroplasty. This suggests that the prosthesis may behave as an unconstrained arthroplasty. This should minimize the stress experienced by the bone-cement interface and may reduce the incidence of loosening. The laxity of the elbows after joint arthroplasty were only slightly greater than normal, possibly explaining the low incidence of prosthesis dislocation which has been observed with clinical use.

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The results of total elbow arthroplasty have been somewhat inconsistent relative to those of other joint replacements, with high rates of loosening with longer periods of follow-up (Garrett et al. 1977, Morrey et al. 1981). Unconstrained total elbows, while having a lower incidence of loosening than hinged implants, have been complicated by postoperative joint instability in nearly a third of patients with some designs (Davis et al. 1982, Lowe et al. 1984, Trancik et al. 1987, Friedman et al. 1989, Weiland et al. 1989, Dennis et al. 1990, Poll et al. 1991, Ruth and Wilde 1992). While unconstrained implants have some intrinsic stability based on the congruency of their articular surfaces (King et al. 1993), the soft tissues, notably the capsule, collateral ligaments, and dynamic muscular supports are considered to be the joint stabilizers, coapting the prosthetic components and maintaining their alignment. This load-sharing may protect against excessive loading of the implant interface and reduce the incidence of aseptic loosening.

In an attempt to address this balance between implant constraint and joint instability, the Norway elbow prosthesis (Figure 1) was developed in 1976 by

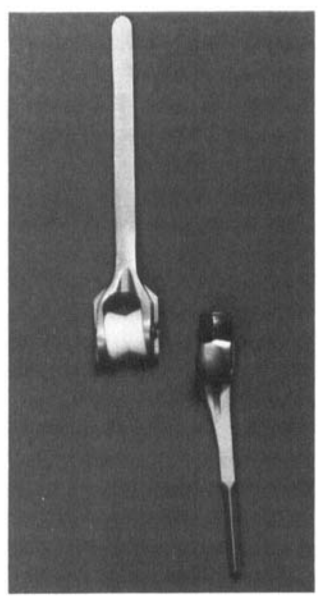


Figure 1. The Norway total elbow system. Note the jaw-like end of the ulnar component which wraps around the humeral bobbin permitting both angular and rotational play, while the prolonged lips reduce the incidence of joint dislocation.

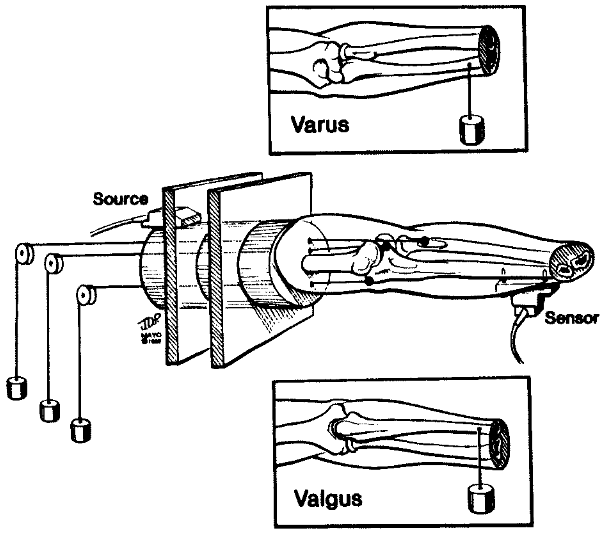


Figure 2. The plexiglass testing apparatus permitted the humerus to be rotated along its longitudinal axis so that valgus and varus forces (see insets) could be applied to the elbow. Simulated muscle-loading was applied to the tendons of the biceps, brachialis, and triceps. The relationship of the source to the sensor was measured using an electromagnetic tracking device.

a Norwegian group of surgeons (Risung 1991). This implant appears to have considerably more intrinsic constraint than other unconstrained prostheses. Clinical results in patients with rheumatoid arthritis have been promising (Risung 1993), which prompted this biomechanical evaluation of the Norway prosthesis in vitro. Our objective was to compare the 3-dimensional kinematics of the Norway implant to the motion patterns of the intact elbow in cadavers. The effect of simulated muscle-loading and the contribution of the radial head to implant stability were also evaluated.

Material and methods

Specimens

6 upper extremities from 6 cadavers were stored at –20 °Celsius and thawed overnight prior to use. The average age of the donors was 66 ± 9 years and the sex distributions were equal. There was no history of musculoskeletal disease and the elbows were radiographically normal. The specimens were amputated through the mid-humeral shaft and disarticulated at the radiocarpal joint, taking care to preserve the distal radioulnar joint. The soft tissue envelope of both the elbow and forearm was left intact as much as possible throughout the study. The muscles and soft tissues were removed from the proximal 8 cm of the mid-humeral diaphysis and polymethylmethacrylate bone cement was used to fix the humerus into a plexiglas sleeve. The musculotendinous junctions of the biceps, triceps, and brachialis muscles were identified and a heavy suture was secured to each tendon. The specimens were mounted in a testing apparatus so that the humerus was parallel to the floor and full motion of the elbow was permitted. Rotation of the humerus along its longitudinal axis permitted valgus and varus loading to be applied across the elbow joint (Figure 2). Pulleys were placed in the testing jig to replicate the normal muscle moment to the arms; 2 cm for the triceps and brachialis, and 4 cm for the biceps (An et al. 1981). Simulated muscle forces were applied to the elbow by loading the biceps and brachialis with 2 kg, and the triceps with 4 kg (O'Driscoll et al. 1992).

A hole was drilled through the ulnar head and a cable was attached to permit loading of the forearm for valgus/varus laxity measurements. The elbow was flexed against gravity by lifting the forearm with the ulnar cable and the force required to flex the arm was measured. Forearm rotation was not constrained. In order to apply a similar magnitude of joint moment to each specimen for the valgus/varus tests, an additional force equivalent to 50 percent of that required to flex the forearm against gravity was applied to the distal ulna. All testing was performed at room temperature and the specimens were kept moist with normal saline irrigation.

Measurement systems

An electromagnetic tracking device (3-Space Tracker, Polhemus Navigational Sciences, Colchester, VT, U.S.A.) was used to measure and record the position and orientation of the elbow. The error of this tracking system is less than 0.5 mm for position and 0.5 degree for orientation (Sidles et al. 1988, Morrey et al. 1991). The electromagnetic source was mounted on the testing apparatus so that it was aligned with, and rotated with, the humeral shaft (Figure 2). The sensor was

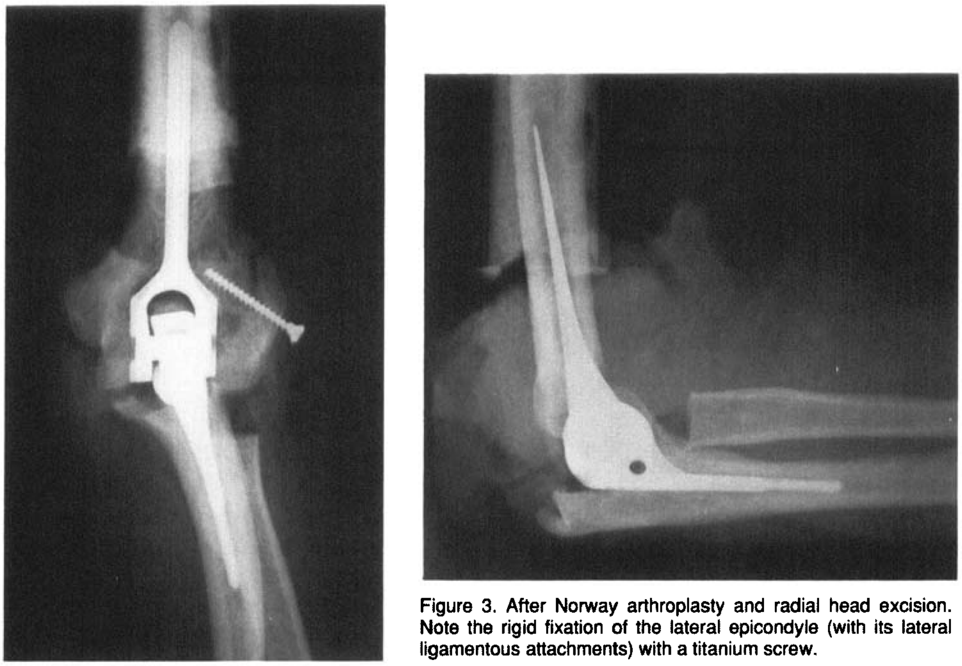


Figure 3. After Norway arthroplasty and radial head excision. Note the rigid fixation of the lateral epicondyle (with its lateral ligamentous attachments) with a titanium screw.

mounted on the distal ulnar diaphysis to monitor the position of the forearm relative to the humerus. A second sensor was mounted on a probe and used to digitize the joint structures of interest.

Data analysis was performed using software specially developed in Minitab (Minitab Inc., State College, PA, U.S.A.; Morrey et al. 1991). Kinematic data from the electromagnetic tracking device were initially transformed from the coordinate system of the source and sensor to an anatomical coordinate system defined for the humerus. Digitization of the trochlea and capitellum with subsequent circle and sphere-fitting of this data permitted the geometric center and anatomical axis of rotation of the distal humerus to be determined. A coordinate system was defined on the basis of this joint axis and the source data was transformed into this anatomical coordinate system. Based on the motion pattern of the intact elbow from 20 to 100 degrees of flexion, an optimized kinematic axis was calculated using principal component analysis. The data were subsequently transformed from the anatomic coordinate system to this optimized kinematic coordinate system. After implantation of the Norway elbow, the data were similarly transformed, using the optimized coordinate system of the intact elbow. This permitted the motion patterns of the total elbow to be compared to those of the intact elbow. The laxity of the intact elbows was also measured in this optimized coordinate system. After implantation laxity data were calculated on the basis of an optimized kinematic coordinate system determined for each total elbow. The laxity data of the intact and total elbows were, therefore, measured in a coordinate system based on the pathway along which each elbow moved, while the position and orientation data were measured in the coordinate system of the intact elbow only.

Prosthetic implantation

To simulate the clinical situation after postoperative soft tissue healing, the surgical procedure was modified to preserve the collateral ligaments of the elbow. An extended lateral approach incorporating a lateral epicondyle osteotomy was used to reattach the lateral ligament complex (Morrey and An 1985). Rigid and reproducible internal fixation of the lateral epicondyle was achieved, using a titanium screw supplemented by bone cement.

Standard implantation techniques and equipment were used by the senior author (FR). After optimal component positioning was achieved, the prosthesis was cemented in place and the joint was reduced. The lateral epicondyle was replaced and the remaining soft tissue envelope was closed in layers. The radial head was not disturbed during the initial implantation. After data collection with the total elbow was completed, the radial head was excised through an anterolateral capsular incision, taking care to preserve the lateral ulnar collateral and radial collateral ligaments. The lateral epicondyle was reattached and a meticulous soft tissue closure was repeated (Figure 3).

Testing protocol

After mounting of the specimen in the testing apparatus, the neutral kinematics of the intact elbow were determined. The elbow was passively extended from a flexed position, at first without loading of the musculotendinous units, and then with the addition of the muscle loads, as previously described. In each case, data were collected in real-time at 60 Hertz for 5 seconds, including yaw (flexion/extension), pitch (valgus/varus), and roll (internal/external rotation). The effect of the titanium implant on the measured joint kinematics was investigated at the beginning of the test protocol by comparing the orientation data with and without the Norway implant taped to the skin of the antecubital fossa.

Valgus-varus laxity was measured as the difference in the orientation of the forearm with the humeral shaft rotated 90 degrees internally and externally from the horizontal or neutral position, as shown in Figure 2. The weight of the forearm, as well as the additional ulnar weight, applied an average moment of 1.1 ± 0.2 N-m across the elbow joint. The coupled forearm rotation measured with valgus-varus loading was defined as rotatory laxity. The influence of muscle loading on joint stability was determined by measuring the joint laxity both with and without the addition of simulated muscle loading.

After the intact elbow was studied to provide normal data, the Norway prosthesis was implanted, as previously described, and the data collection was repeated. Finally, the radial head was resected to determine the importance of this structure in stabilizing the arthroplasty.

The valgus-varus and rotatory orientations and laxity of the forearm were determined at 30, 65, and 100 degrees, with simulated muscle loading. The maximal valgus-varus and rotatory laxity limits were calculated over a range of motion from 20 to 100 degrees both with and without muscle loading. These measurements and comparisons were determined for each condition of the elbow; intact, and implants before and after removal of the radial head.

The position and orientation of the kinematic axes of the intact elbows were compared to these of the implants as a measure of how closely the implant mimicked normal motion. The position and orientation of the axis of the trochlear groove were compared to the groove in the humeral component and the axis of the central ridge of the ulnar component was compared to the central ridge of the semilunar notch of the ulna as an indication of the surgical accuracy of component implantation.

Statistical analysis was performed using SAS (Version 6.06, SAS Institute Inc., Cary, NC, U.S.A.). A two-way repeated measures analysis of variance was used to evaluate the effect of the elbow condition and either joint angle or muscle loading on the outcome variables of interest. The Student-Neuman-Keuls multiple comparison procedure was used to further characterize the groups with statistical differences. One-way repeated measurements analysis of variance or paired *t*-tests were used to compare groups with significant variable interactions in the two-way tests. Paired *t*-tests were performed to compare the position and orientation of the kinematic axes and those of the humeral and ulnar components. A Bonferroni correction factor was used to adjust the significance level of the paired *t*-tests.

Results

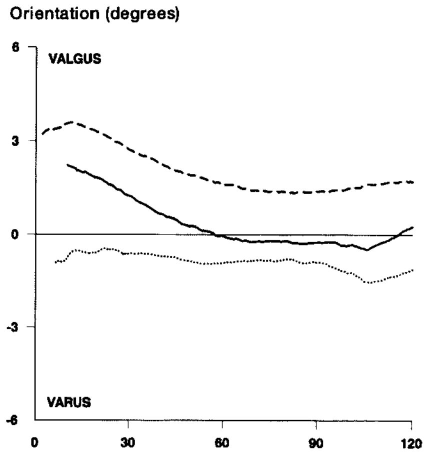
The valgus-varus and rotatory data were highly reproducible after external placement of the Norway prosthesis on the elbow, suggesting minimal interference of the titanium implant with the accuracy of the measurement device. The mean difference in valgus-varus and rotatory orientations averaged 0.0 ± 0.1 (*P* 0.2) and 0.0 ± 0.1 (*P* 0.08) degrees, while the maximal differences were 0.3 and 0.2 degrees, respectively. These small differences were not statistically significant, using a paired *t*-test.

The motion patterns of the implant were similar to those of the intact specimens (Figure 4). The orientation of the implants tended to be marginally varus relative to normal and both the intact and implant elbows were somewhat more valgus in extension and became more varus with flexion (Table 1). After implantation, the forearms were more externally rotated than normal throughout the range of flexion-extension. Excision of the radial head did not change the neutral kinematics of the implant.

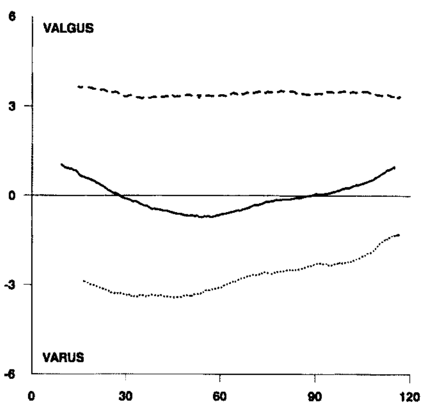
The valgus-varus laxity of the normal elbows was small, the average limits being 5.6 degrees for the intact specimens, increasing to 7.9 and 9.6 degrees for the implant elbow with and without the radial head, respectively (Table 2). Muscle loading stabilized the elbows, reducing the valgus-varus laxity limit to 3.3, 5.8, and 6.8 degrees. The valgus-varus laxity of the total elbows were significantly greater than normal at 30, 65, and 100 degrees of flexion (Table 3). There was no effect of flexion angle on valgus- varus laxity.

The rotatory laxity limit of the elbows was not different after implantation (Table 2). Muscle loading reduced the magnitude of coupled forearm rotation for all elbow conditions. Implantation increased the rotatory laxity at 30, 65, and 100 degrees of flexion (Table 3). Coupled rotation did not differ from the intact

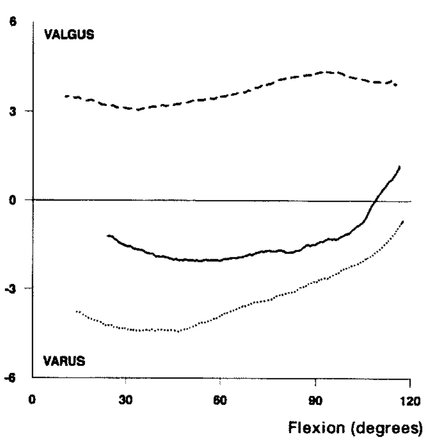
Figure 4. Kinematics of intact and implant elbows.
- - - valgus; —— neutral ; · · · · · varus



Intact—sample curves depicting the motion pattern of an intact elbow with and without valgus-varus loading.



Norway elbow—sample curves after implant arthroplasty with and without valgus-varus loading. Note the motion patterns similar to those in the intact elbow. However, an increase in valgus-varus laxity is noted throughout the range of flexion.



Norway elbow after radial head excision—sample curves demonstrate an increase in the valgus-varus laxity after radial head excision.

Table 1. Orientation of the elbows. Positive values represent valgus or internal rotation in degrees. Mean *SD*

	Intact	Radial head	No radial head	*P*-value (implant)
Valgus/varus orientation				
30°	1.1 *0.7*	0.7 *2.0*	0.8 *2.1*	
65°	0.4 *0.6*	0.4 *1.0*	0.2 *1.2*	0.4
100°	0.3 *0.4*	–0.5 *0.48*	–0.5 *1.3*	
P-value (angle)		0.03		
Rotational orientation				
30°	–0.9 *2.2*	2.3 *2.0*	2.4 *2.3*	
65°	–0.9 *0.8*	1.2 *1.4*	1.6 *1.8*	0.0001
100°	–0.4 *0.9*	0.8 *1.0*	0.7 *0.95*	
P-value (angle)		0.09		

Table 2. Laxity limit (degrees) of the elbows. Mean *SD*

	Intact	Radial head	No radial head	*P*-value (implant)
Valgus/varus laxity limit				
1/2 kg	5.6 *1.9*	7.9 *1.7*	9.6 *1.4*	
2/4 kg	3.3 *0.7*	5.8 *1.8*	6.8 *2.1*	0.0001
P-value (angle)		0.0001		
Rotational laxity limit				
1/2 kg	3.9 *0.9*	3.5 *1.7*	4.3 *1.5*	
2/4 kg	2.5 *0.7*	3.1 *1.3*	3.6 *1.4*	0.1
P-value (angle)		0.02		

Table 3. Laxity (degrees) of the elbows. Mean *SD*

	Intact	Radial head	No radial head	*P*-value (implant)
Valgus/varus laxity				
30°	3.1 *0.6*	5.1 *2.1*	6.2 *2.2*	
65°	2.1 *0.4*	5.0 *1.9*	5.9 *2.3*	0.0001
100°	2.1 *0.5*	4.8 *2.0*	5.6 *2.4*	
P-value (angle)		0.3		
Rotational laxity				
30°	2.3 *0.7*	2.6 *1.4*	3.0 *1.6*	
65°	1.8 *0.5*	2.3 *1.3*	2.6 *1.7*	0.001
100°	0.9 *0.5*	2.0 *1.2*	2.5 *1.1*	
P-value (angle)		0.06		

elbows; however, after excision of the radial head, ulnar rotation became greater than normal.

The kinematic axes of the Norway elbows were in 1.7 degrees of external rotation relative to the intact specimens (Table 4). The orientation of the humeral components was similar to the trochlear groove of the humerus. The positions of the humeral components were similar in their proximal-distal and anterior-pos-

Table 4. Orientation and position of the total elbows relative to normal. Positive values represent valgus or internal rotation orientation and proximal, posterior, or lateral positions. Mean *SD*

	Kinematic axes		Component axes Humerus		Component axes Ulna	
Orientation (degrees)						
Valgus-varus	2.3	*4.0*	0.9	*2.8*	–11.3	*11.0*
P-value	0.2		0.5		0.05	
Int-ext rotation	–1.7	*1.5*	–3.2	*5.3*	3.4	*4.2*
P-value	0.04		0.2		0.1	
Position (mm)						
Proximal-distal	1.1	*3.5*	–0.5	*1.2*	–1.3	*1.7*
P-value	0.5		0.6		0.1	
Posterior-anterior	1.3	*1.4*	0.2	*0.8*	0.1	*0.2*
P-value	0.07		0.6		0.6	
Lateral-medial	–		2.9	*1.6*	0.0	*0.1*
P-value			0.003		0.4	

terior positioning, however, they were placed an average of 2.9 ± 1.6 mm lateral to the normal trochlear groove. The positions of the anatomic axis of the ulnar components were similar to that of the greater sigmoid notch of the ulna, however, their orientations tended to be highly variable, as reflected by their large standard deviations.

Discussion

The Norway arthroplasty closely replicated the motion patterns of the normal elbow joint, however, some subtle differences were noted. The kinematic axes of the Norway elbows were orientated in about 2 degrees of external rotation relative to normal. The humeral components were implanted in about 3 degrees of external rotation which may explain the observed external rotation of the ulna relative to the humerus after arthroplasty. Thus, surgical error, rather than a deficiency in prosthesis design, is probably responsible for the measured differences in motion patterns.

The orientations of the ulnar components were highly variable. This reflects the difficulty in defining reproducible anatomic landmarks of the ulna at the time of prosthesis implantation. In clinical practice, the distorted anatomy makes the orientation of the components more difficult than that of our normal joints. One would expect that even more variability in motion patterns might be observed clinically with this implant than was appreciated in this study. Close attention to detail, while orienting both the humeral and ulnar components, is essential to reproduce the kinematics of the normal elbow.

It is not clear why the valgus-varus laxity of the Norway implants was greater than normal. Assuming that the capsulo-ligamentous material properties were unchanged, and the soft tissue envelope was adequately reconstructed, this increase in laxity can only be explained by a failure to restore the normal distance between the collateral ligament insertions. While abnormal positioning of the humeral and/or ulnar components may be responsible for this abnormal ligament pretensioning, only the lateral positioning of the humeral components differed from normal. The humeral components were placed about 3 mm lateral to the trochlear groove of the humerus, which would tend to increase the medial collateral ligament tension and relax the lateral collateral tension. The abnormal valgus-varus and rotatory orientation of the humeral and ulnar components, as previously discussed, will also generate an imbalance in the collateral ligament tension and may also explain the increase in joint laxity after Norway arthroplasty.

Another possible explanation for the observed increase in laxity after joint replacement is found in the experimental model. Although the radial collateral and lateral ulnar collateral ligaments were reconstituted using a lateral epicondyle osteotomy, the joint capsule and muscles which were stripped from the distal humerus and proximal ulna could not be returned to their preoperative status and may be responsible for some of the measured increase in joint laxity. Even though the implant was stabilized by muscle loading, the joint laxity was not restored to normal. This suggests that while muscle loading may provide important dynamic secondary restraint, the static capsulo-ligamentous supports are probably the primary stabilizers of this implant.

The implication of an increase in the valgus-varus laxity after implantation of the Norway elbow is unknown. The articulation of the Norway prosthesis, by virtue of its design, permits much greater valgus-varus laxity than was measured in this study. This suggests that the Norway elbow was constrained by the capsule, ligaments, and muscles rather than the implant itself. The prosthesis, therefore, functioned as an unconstrained arthroplasty when implanted in vitro. This should minimize the stresses on the bone-cement interface in vivo and reduce the incidence of loosening relative to more constrained implants.

Excision of the radial head increased the valgus-varus laxity of the elbow after implantation of the Norway prosthesis. This is in contrast to studies of normal elbows where excision of the radial head has been demonstrated to have little effect on joint stability (Hotchkiss and Weiland 1987, Morrey et al. 1991). The decrease in stability could also be explained by a deterioration in the quality of the soft tissues with repeated testing or a less adequate repair of the soft tissue. Excision of the radial head and data collection, however, was completed within 30 minutes, suggesting that little change in the properties of the soft tissue envelope should have occurred. The radial head was easily removed without damaging the collateral ligaments due to the nature of the surgical approach. These 2 factors suggest that the measured increase in valgus-varus laxity after excision of the radial head are probably real and not an artifact of the experimental design. The decrease in stability after excision of the radial head is probably best explained by a decrease in the soft tissue tension of the radial collateral ligaments.

This investigation demonstrates that the Norway prosthesis can closely replicate normal elbow kinematics, provided satisfactory component positioning and orientation is achieved which, in cases of advanced joint destruction, may prove difficult. The stability of the joint would appear to be adequate for clinical purposes if the collateral ligaments are structurally sound and their tension appropriately balanced. The dynamic effect of muscles which pass across the elbow should provide additional stability to the elbow arthroplasty. Preservation of the radial head, where possible, may be worthwhile to improve the stability of the elbow after joint replacement.

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