

Telemeterized In Vivo Hip Joint Force Data: A Report on Two Patients After Total Hip Surgery

G. M. Kotzar, D. T. Davy, V. M. Goldberg, K. G. Heiple, J. Berilla, K. G. Heiple, Jr.,
*R. H. Brown, and †A. H. Burstein

*Orthopaedic Engineering Laboratory, Departments of Orthopaedic Surgery and Mechanical and Aerospace Engineering, Case Western Reserve University, Cleveland, Ohio; *Division of Surgical Research, St. Luke's Hospital, Cleveland, Ohio; and †Department of Biomechanics, Hospital for Special Surgery, New York, New York, U.S.A.*

Summary: Two telemeterized femoral components were implanted in two patients as part of normal total hip replacement procedures. The two components were instrumented to measure the three force components directed along: (a) the neck axis, (b) transverse to the neck axis and in the plane of the prosthesis, and (c) transverse to the neck axis and perpendicular to the plane of the prosthesis. Data were collected at multiple sessions during the early postoperative period for a number of standard activities, including gait, stair climbing, rising from a chair, single leg stance, double leg stance, ipsilateral and contralateral straight leg lifts while supine, ipsilateral flexion and extension while standing, and ipsilateral abduction while standing and lying on the contralateral side. These data are summarized and compared with the published results from analytic studies and with the results from previous studies using instrumented femoral components. Peak loads for gait during the period of study were roughly 2.7 body weights (BW) when the patients walked at their normal pace. Contact forces at the hip during stationary single leg stance approximated the peak loads during gait with values ranging from 2.1 to 2.8 BW. The highest forces recorded reached values approaching 5.5 BW and occurred during periods of instability while the patient engaged in stationary single leg stance. Our in vivo data indicate that forces generated during the above activities increase in magnitude quite rapidly during the early postoperative period and that during this period the patients have the ability to perform the activities of daily living without generating the high amplitude joint contact forces suggested by the results of dynamic studies. Joint contact forces during gait were found to depend on speed, but the high absolute magnitudes predicted by model studies were not supported by the in vivo data. **Key Words:** Telemetry—Joint contact force—In vivo forces—Hip joint.

The question of what loads are carried across the hip joint is of interest to the clinician and the researcher alike. It has not been possible to obtain direct in vivo force measurements in the intact hip,

because no acceptable invasive measures have been developed yet. However, the advent of hip arthroplasty created the potential for measuring in vivo loads across prosthetic joints by instrumenting the prosthesis.

The first successful attempt to use prosthetic joints for force measurement was the work of Rydell, who implanted two instrumented femoral components in a hemiarthroplasty procedure (21). Ry-

Received August 9, 1989; accepted November 1, 1990.
Address correspondence and reprint requests to Dr. Dwight T. Davy, Department of Mechanical and Aerospace Engineering, Glennan Building, Room 615A, Case Western Reserve University, Cleveland, OH 44106, U.S.A.

dell's pioneering work established nominal load values at 6 months postsurgery for a variety of activities. Because the access of the device was via percutaneous leads, however, it was used only for a data collection series at a single time.

The potential for using telemeterized devices was also recognized fairly early. In work related to the hip joint, our laboratory instrumented femoral hip nails and obtained loads carried by them in a variety of early postoperative activities (4). English and Kilvington (11) reported on measurements in one patient using a telemeterized system whereby they measured the force along a single "effective" axis of an instrumented femoral component in a total joint prosthesis. English and Kilvington's work indicated force amplitudes during gait comparable with those obtained by Rydell, but the only data reported by English and Kilvington were at 42 days postsurgery. A different approach to *in vivo* force measurements was used by Hodge et al. (14), who instrumented a hemiarthroplasty component with multiple local pressure transducers. This device was implanted in one patient and produced information concerning local contact pressures on the prosthesis, but it did not determine total load data.

The alternative means to *in vivo* force measurements has been to use models of musculoskeletal dynamics to predict forces. Such models have been used to estimate the internal forces in the joint based on the measurement of external kinematics and kinetics, the estimation of inertial and geometric parameters associated with the musculoskeletal system model, and the solution of the so-called inverse dynamics problem to determine resultant force and torque histories at the joints (8). These resultants are then distributed among the various load-carrying elements via one of a variety of modeling techniques to predict the forces in muscles, ligaments, other tissues, and the joint contact surfaces.

To solve the indeterminate problem of force sharing in a redundant system, several researchers (15,16,18,20,22) grouped the muscles into coactive sets. Others have adopted methods from optimization theory to arrive at unique solutions for muscle forces and joint contact forces (2,6,8,13,20). A broad range of predicted hip joint contact force values have resulted from the various models. In a model of single leg stance, forces have ranged from as low as 1.8 body weights (BW) (15) to six BW (22). In normal gait, predictions have ranged from approximately three BW up to eight BW (6,8,9,13,

17,20). For activities other than gait, large hip joint contact forces have also been predicted. Paul (18) predicted forces in excess of seven BW for stair ascent and descent. Crowninshield et al. (8) predicted forces in excess of eight BW for stair ascent, and values in excess of three BW for rising from a chair. The predicted forces for these two activities are of particular interest because they have large out-of-plane components compared with values for gait.

In the present report, *in vivo* force data from the early postoperative period for two subjects are presented and discussed in light of previous *in vivo* data and model studies. We report telemeterized total hip replacement prosthesis data for one patient over 31 days and for a second patient over 58 days. Data collection was terminated shortly after these times due to technical difficulties with the electronics.

METHODS

The basic design configuration has been previously described (10) as has the instrumentation and telemetry electronics (5). The femoral component of a total hip replacement prosthesis is instrumented to act as a three degree-of-freedom load cell capable of measuring the three load components directed along the axis of the neck (AX), the load component transverse to the neck axis and in the plane of the prosthesis (IP), and the transverse load component perpendicular to the plane of the prosthesis (OP). Semiconductor strain gauges are mounted inside the hollow neck, and temperature compensation is provided by wiring them into Wheatstone Bridge configurations with two active arms for the two transverse bridges (IP and OP) and four active arms for the axial bridge (AX). These three channels are driven by independent electronic oscillators, the amplitudes of which are modulated by the applied load. A fourth unmodulated channel is used as a reference (5,10). The four channels are multiplexed and transmitted as a single FM signal. An FM receiver is used to receive the signal, which is separated into individual channels by demodulation.

Each prosthesis was calibrated completely under computer control in two steps. The instrumentation electronics were first calibrated as a function of excitation voltage and temperature. To control the temperature to the required level of accuracy, a special loading fixture was constructed that was ca-

pable of maintaining a desired temperature to $\pm 0.01^\circ\text{C}$. The temperature range over which the devices were calibrated was $34\text{--}40^\circ\text{C}$ in one-half degree increments to allow for normal biologic temperature variation as a function of activity. The prostheses were calibrated for voltage over the normal expected range for the batteries used: $3.3\text{--}3.6\text{ V}$ direct current in 0.050-V increments. These voltages were held to $\pm 0.001\text{ V}$ direct current. This allowed the determination of the temperature and voltage dependence of the unmodulated amplitude and frequency of each of the four data channel oscillators.

The second step calibrated the output of each channel as a function of the three parameters: load, voltage, and temperature. The protocol used for voltage and temperature was the same as in the previous step. The load calibration was performed over two different ranges: $0\text{--}3560\text{ N}$ in 445-N increments for the axial channel (AX) and $-1780\text{--}1780\text{ N}$ in 223-N increments for the two transverse channels (IP and OP). The loads were held to an accuracy of $\pm 4.5\text{ N}$. The full range of loads was applied at every temperature and voltage increment for all three channels. The data for the AX, IP, and OP channels at 37°C are shown in Figs. 1, 2, and 3, respectively. Shown on each plot are the primary responses, as well as the crosstalk, resulting from loading along the secondary axes.

The load data was reduced using a second-order multivariable linear regression resulting in a fit to the data with a 95% confidence interval of $\pm 2.5\%$ of full scale.

Two prostheses have been implanted. The first prosthesis was implanted in a 67-year-old woman in June 1984. A second prosthesis was implanted in a 72-year-old man in September 1987. The implant was used for revising a failed uncemented bipolar prosthesis for the first patient, and for primary osteoarthritis in the second patient. Patient data are given in Table 1. Data were obtained for a series of sessions postoperatively for each of the two subjects beginning 3 days after surgery. In each session, the implant was activated, data were recorded continuously for a minimum of 40 minutes, and a set of activities was dictated by the normal postoperative patient protocols. The activities of interest for one or both patients, included, among others, (a) supine ipsi- and contralateral straight leg lifts; (b) double leg stance; (c) ipsilateral single leg stance; (d) flexion and extension of the ipsilateral limb; (e) ipsilateral abduction in the standing position; (f) ipsilateral abduction while lying on the contralateral side; (g) gait with one or two crutches, a cane, or between parallel bars and free gait; (h) stair climbing using crutches, a cane, or the bannister, and without support; (i) rising from a 44.5-cm high chair.

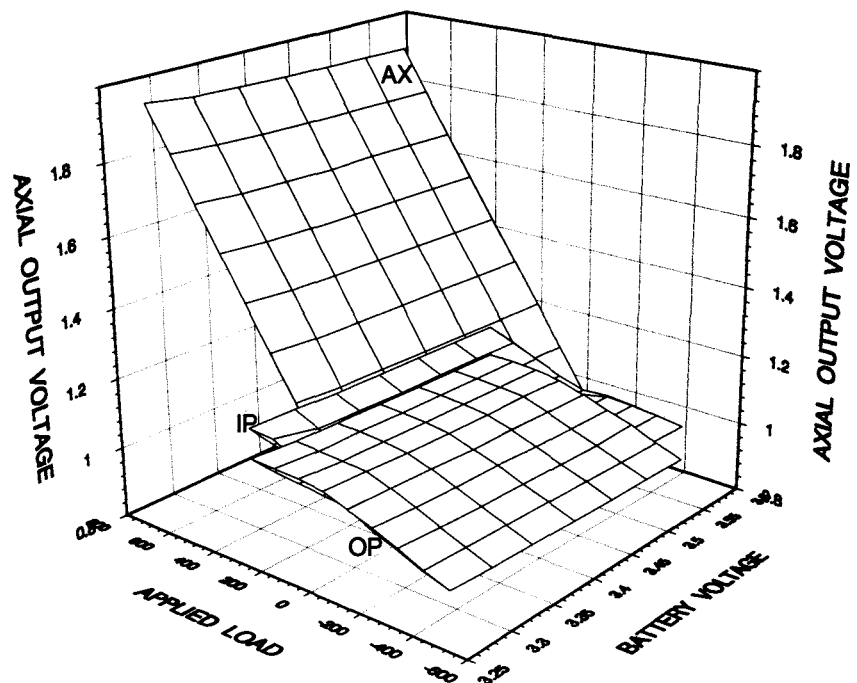


FIG. 1. Plot of the 37°C calibration data for the axial (AX) channel output as a function of battery voltage and applied load. The response for each applied load direction is plotted separately.

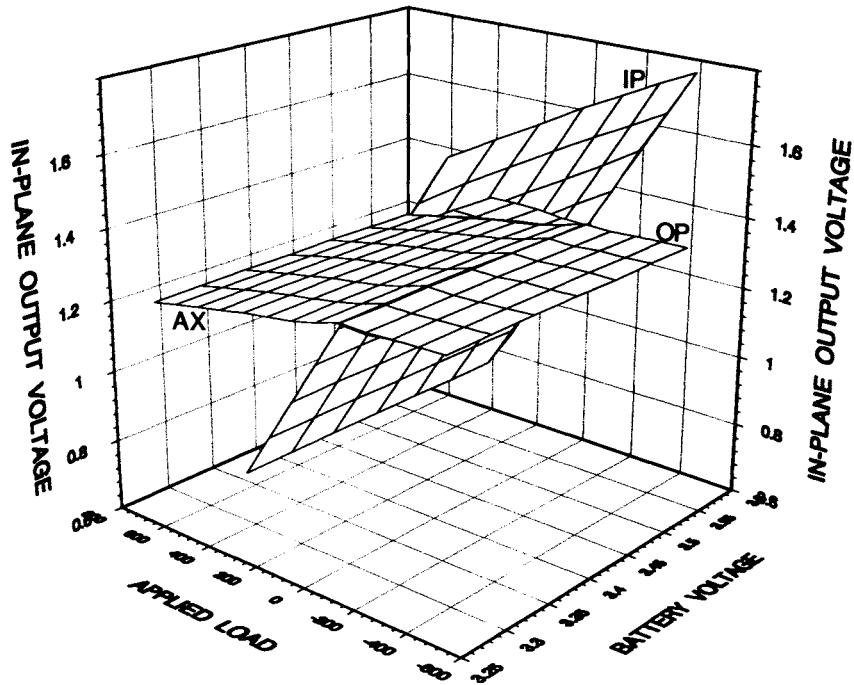


FIG. 2. Plot of the 37°C calibration data for the in-plane bending (IPB) channel output as a function of battery voltage and applied load. The response for each load direction is plotted separately.

Data were obtained and reduced to records of the individual force components AX, IP, and OP. Also, the resultant force R, and angles of orientation of the resultant force, and were calculated. The definitions for R, θ , and ϕ in terms of the components AX, IP, and OP are given in Fig. 4.

RESULTS

Force histories for stationary single leg stance for patient 1 at 31 days and patient 2 at 35 days post-surgery are shown in Figs. 5 and 6, respectively. From Table 2 it can be seen that the time-averaged

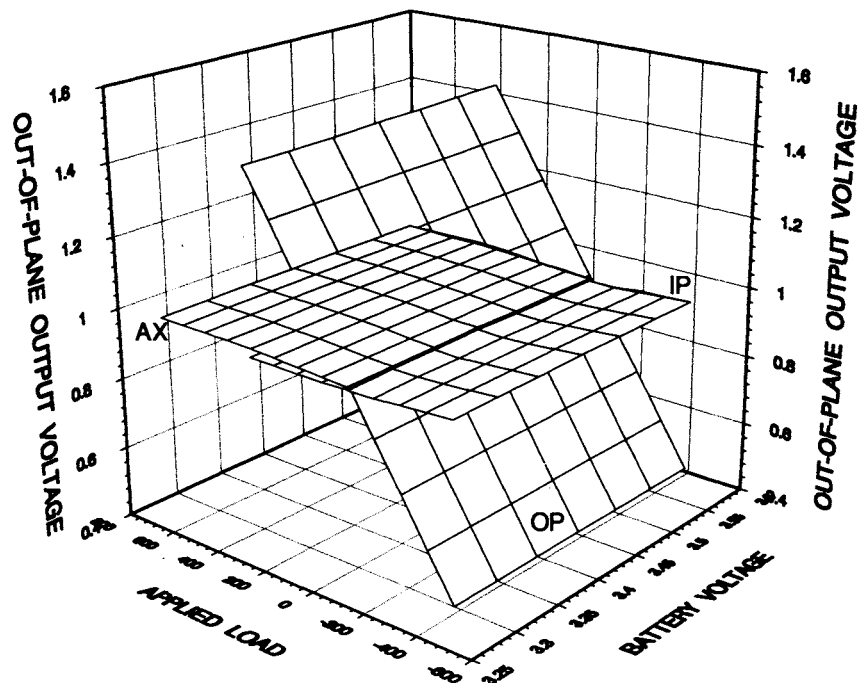


FIG. 3. Plot of the 37°C calibration data for the out-of-plane bending (OPB) channel output as a function of battery voltage and applied load. The response for each load direction is plotted separately.

TABLE 1. Telemeterized hip patient data

Data	Patient 1	Patient 2
Sex	F	M
Age (yr)	67	72
Height (m)	1.56	1.70
Weight (n)	556	620
Procedure	Revision	Primary

amplitudes (referenced to BW) were similar between the two subjects for single leg stance. However, patient 2 showed a greater variability in joint contact force over time; e.g., in Fig. 3 the instantaneous peak force reached a magnitude of 5.4 BW. Patient 1 performed single leg stance with one of her hands in the hand of an attendant for stability. Patient 2 used no support while performing single leg stance and often exerted an obvious effort to maintain his balance. As summarized in Table 2, the time-averaged amplitudes of the resultant joint contact forces, excluding the large-magnitude transients, as noted, were found to be between two and three BW for both patients, and were essentially independent of the time after surgery at which the data were collected.

During the postoperative period, gait data were collected under varying circumstances, depending on the capabilities of the individual at that time. For the earlier data sessions, unassisted gait was approximated by having the patient walk between parallel bars that had been adjusted to the proper height. The patients were instructed to make as little use of the parallel bars as possible, guided by their own comfort level. They were allowed to walk at the pace at which they felt comfortable. When the attending physician felt that the patient was ready, data for completely unassisted gait were collected. Only patient 2 provided data for free gait.

TABLE 2. Time-averaged forces for stationary single leg stance

Patient	Days postsurgery	Resultant force (\times BW)		
		Average ^a	No. ^b	Range ^c
1	31	2.1	1	2.0-2.3
2	23	2.8	2	2.7-4.5 ^d
	35	2.7	1	2.2-5.4 ^d
	44	2.6	4	2.4-3.5
	58	2.6	4	2.4-5.0 ^d

^a Individual loads computed by integrating over time for each single stance event.

^b Number of events.

^c Instantaneous peak loads.

^d Single isolated instantaneous value excluded from time averaging.

Shown in Figs. 7 and 8 are typical gait data for both subjects during the final steps of gait between parallel bars, as described earlier. The times at which these data were collected were 31 days post-surgery for patient 1 and 44 days postsurgery for patient 2. At these times both patients walked with

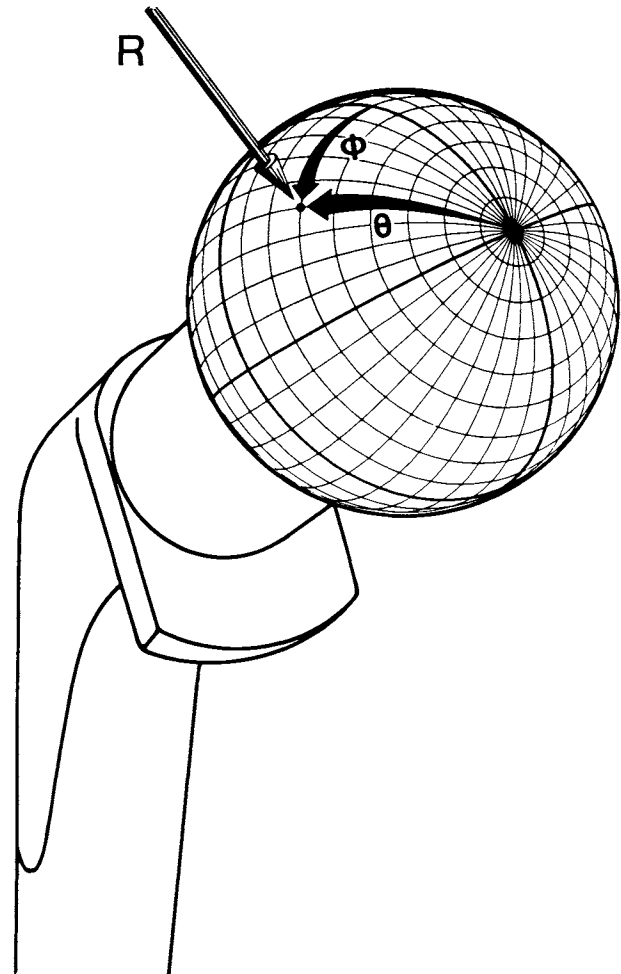


FIG. 4. Definitions of the resultant force and orientation angles. The resultant (R) defines the magnitude of the effective joint contact force. The orientation angles Θ and Φ determine the effective contact point and orientation of the contact force on the ball of the prosthesis. The cone angle (Θ) defines orientation with respect to the axis of the neck of the prosthesis, and is the angle between the resultant force and the neck of the prosthesis. It is defined by $\cosine \Theta = AX/R$. A value of $\Theta = 0$ would indicate a force on the apex of the ball directed along the neck axis. The polar angle (Φ) defines orientation with respect to the plane of the prosthesis; it defines the angle between the force and the midplane of the prosthesis. It is given by $\tangent \Phi = OP/IP$. A value of $\Phi = 0$ would indicate a force acting in the plane of the prosthesis on the superior aspect. For a prosthesis implanted in the right hip, a negative value of Φ would indicate a posteriorly directed resultant and an anteriorly located contact point. The point shown corresponds to values of $\Theta = 60^\circ$ and $\Phi = -50^\circ$.

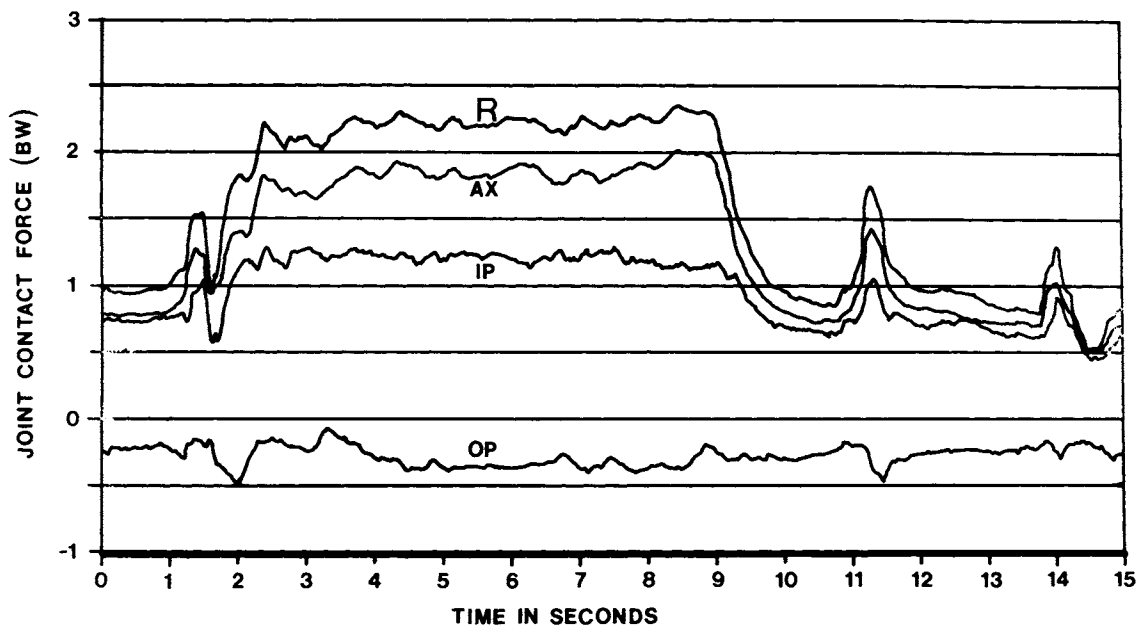


FIG. 5. Force history for single leg stance for patient 1 at 31 days postsurgery. Hips were approximately horizontal in the frontal view.

no apparent support from the bars, merely sliding their hands along them. At 44 days, patient 2 also walked with no support. The data were compared for him when walking unassisted and walking be-

tween parallel bars at comparable speeds. Using Student's *t* test for differences between means with like variances, no statistically significant differences in average force magnitudes were found. We

TABLE 3. Average peak forces for gait activities

Patient	Days postsurgery	Type of support	Resultant force (\times BW)			
			Average ^a	SD	Steps	Range ^b
1	31	Crutches	2.8	0.1	32	2.6-3.1
		Parallel bars	2.4	0.1	27	2.3-2.8
2	14	Crutches	1.0	0.3	21	0.6-1.7
		Parallel bars	1.0	0.4	14	1.1-2.4
2	23	Crutches	1.2	0.2	22	0.8-1.4
		Crutch (ipsilateral)	2.0	0.3	7	1.5-2.4
		Crutch (contralateral)	1.6	0.3	5	1.4-2.2
		Cane	1.9	0.4	6	1.5-2.7
		Parallel bars	2.2	0.2	21	1.9-2.5
		Free	2.0	0.2	12	1.8-2.4
		Free (slow)	2.4	0.2	7	2.2-2.8
2	35	Crutches	2.0	0.3	13	1.5-2.5
		Free (normal)	2.7	0.3	10	2.4-3.3
		Free (slow)	2.4	0.2	7	2.2-2.8
		Crutch (contralateral)	2.4	0.3	23	2.0-2.9
2	44	Cane (contralateral)	2.3	0.2	14	1.9-2.9
		Parallel bars	2.6	0.3	10	2.2-3.1
		Free (normal)	2.7	0.3	10	2.2-3.2
		Free (slow)	2.3	0.2	30	1.8-2.6
		Free (very quick)	3.6	0.2	8	3.2-3.9
2	58	Cane	2.6	0.3	13	2.2-3.2
		Free (normal)	2.5	0.3	13	2.3-3.3
		Free (slow)	2.4	0.1	5	2.3-2.5
		Free (quick)	2.8	0.2	10	2.7-3.1
		Free (very quick)	3.6	0.2	8	3.2-3.9

^a Peak force attained on each step over total steps.

^b Range is of peak forces over total steps.

have included the force histories for parallel bars in Figs. 7 and 8 in order to present the most direct comparisons between subjects. It is of interest to note from Figs. 7 and 8 that the hip joint is unloaded only for a relatively short period of the total gait cycle. Over all gait data for our two patients, the unloaded period was 15–20% of the total gait cycle.

As can be seen from these records, the cadences for the two subjects were quite different. Speeds were also markedly different: patient 1 walked at a speed of 0.43–0.50 m/second; patient 2 walked at a speed of 1.0–1.1 m/second. These values for speed are average values based on the time to traverse the known length of the walkway. The characteristics for the gait of patient 2 as shown are representative of his gait at both earlier and later times postsurgery. Patient 2 tended to start moving quickly and then settle into a pace, which typically produced the highest force readings during the first one or two steps of each sequence. The next three or four steps produced fairly constant but lower peak loads, with the final step or two displaying a marked reduction in amplitude. A summary of average values of peak loads during gait under a variety of circumstances is given in Table 3. As was found for the first subject, patient 2 showed an early plateauing of forces after a fairly rapid rise in peak force amplitudes during gait over the first 3 weeks.

Included in Table 3 are peak resultant force averages over a number of steps for patient walking at different speeds at several times postsurgery. The gait speeds were established by asking the patient to walk at his normal gait, at a slow gait, and at a fast gait limited by his comfort level. It can be seen from the data that the trend is toward higher amplitudes with higher walking speeds, and higher amplitudes as a function of time postsurgery. The general speeds for these sessions ranged from about .9 m/second for slow walking to nominally 1.1 m/second for normal gait and 1.3 m/second for fast gait.

Shown in Fig. 10 are force data for a sequence of stair climbing by patient 2. The sequence shown is for stair climbing without aid. As can be seen, the force amplitudes are relatively low. In comparison with normal gait there is a significant change of angle of the resultant with respect to the plane of the prosthesis. A comparison of force amplitudes for various methods of walking upstairs as a function of time postsurgery is given in Table 4. It can be seen that the force amplitudes are modulated by the use of aids, but the peak forces remained at relatively low values even when the subject was ascending the stairs without aids.

In Fig. 11 are shown data for patient 2 while rising from a chair without the use of arm support. Aver-

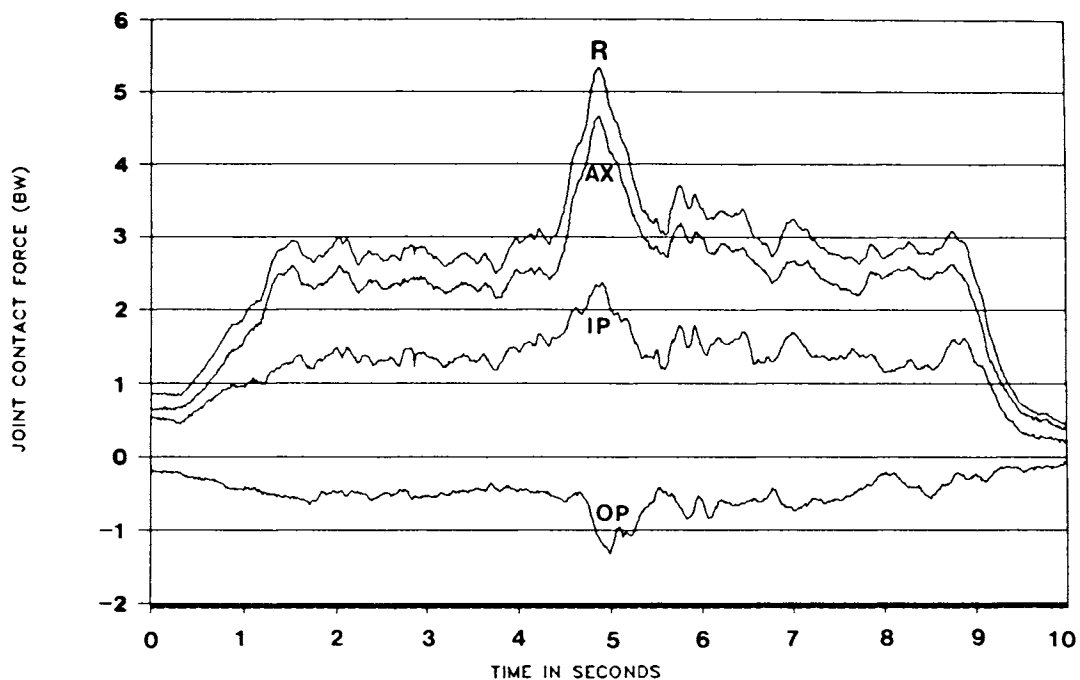


FIG. 6. Force history for single leg stance for patient 2 at 35 days postsurgery. Hips were approximately horizontal in the frontal view.

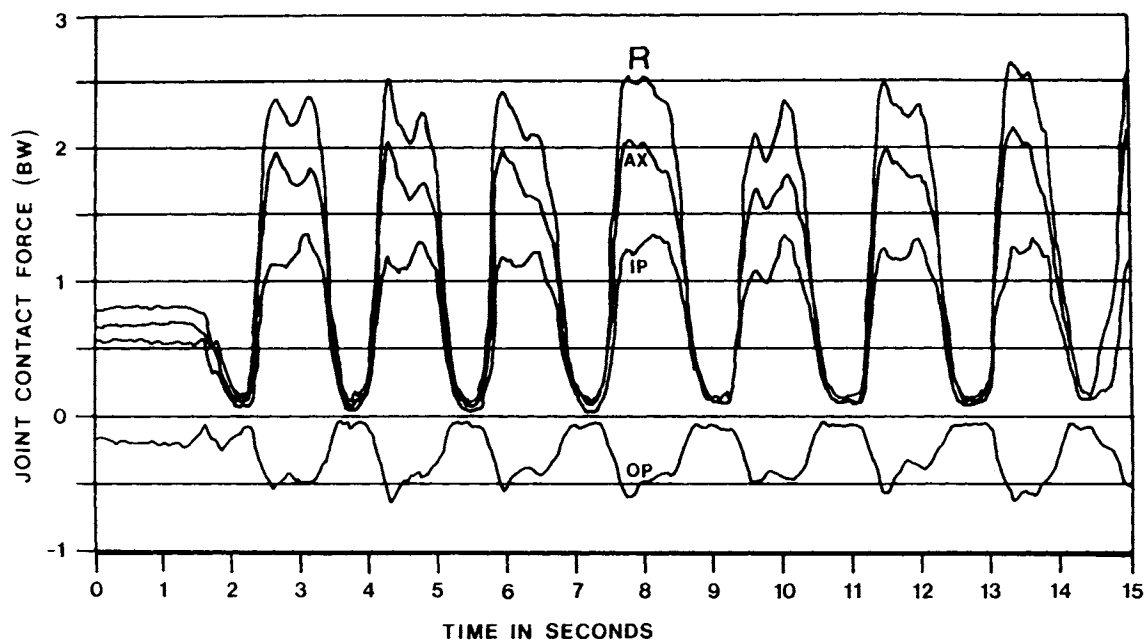


FIG. 7. Force history for gait for patient 1 at 31 days postsurgery. Subject walked between parallel bars while sliding hands along bars. Two-legged support was present for 27% of the stride cycle, and the stride length was 0.87 m.

age peak forces over several sequences of rising from a chair for the two subjects are shown in Table 5. As for the case of stair climbing, force amplitudes could be modulated by use of aids; however, in rising without the use of the chair arm, patient 2 produced total contact forces of less than 1.5 BW.

In addition to the data presented here, data were obtained for a variety of additional activities for both patients. These included activities such as lying down and getting up from a bed, and doing a variety of leg maneuvers on the ipsilateral side, including extension-flexion and abduction maneuvers. Typical force values for various of these activities are given in Table 6.

TABLE 4. Average peak forces for stair climbing

Patient	Days postsurgery	Type of support	Resultant force (\times BW)			
			Average ^a	SD	Steps	Range ^b
1	31	Crutches	1.5	0.4	12	1.0-2.3
		Bannister	2.3	0.2	15	2.1-2.6
2	14	Crutches	1.0	0.3	10	0.5-1.8
		Crutch	1.5	0.4	12	1.0-2.3
2	35	Bannister	2.4	0.3	14	1.9-3.0
		Bannister	2.3	0.2	19	1.8-2.8
2	58	Cane	2.2	0.4	22	1.5-2.8
		Bannister	2.5	0.3	6	2.1-2.8
		Free	2.6	0.1	8	2.1-2.8

^a Peak force attained on each step for total steps.

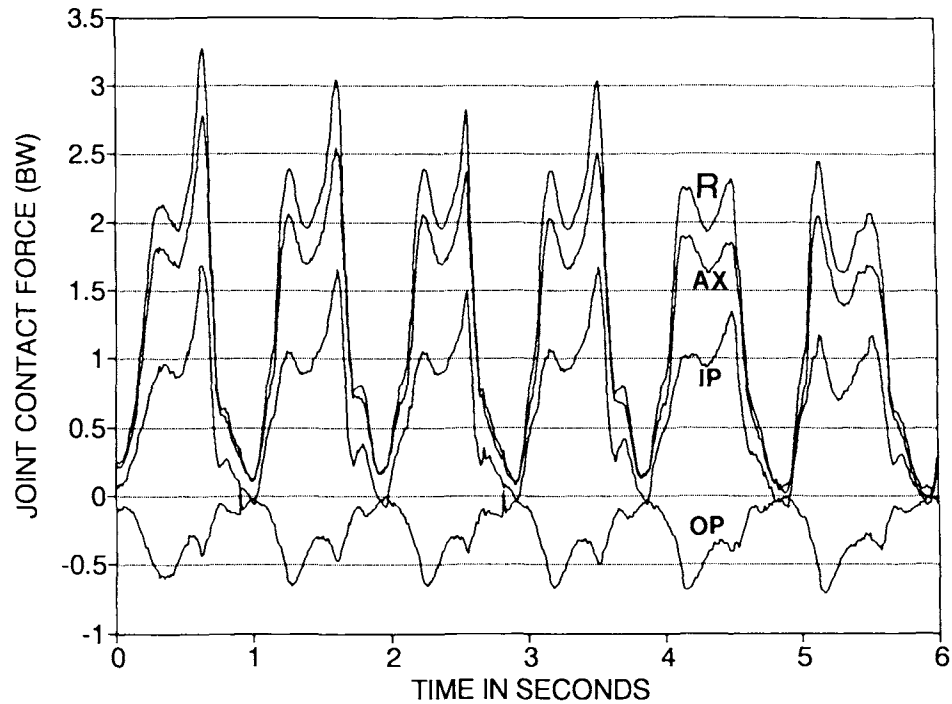
^b Range is of peak forces over total steps.

DISCUSSION

The data from the two patients we have studied show that during the early postoperative period total hip arthroplasty patients can do a number of familiar activities with relatively low joint contact forces, as compared with results of external motion studies of normal subjects. For example, these studies, which combine external measurements and analytical models, have predicted maximum hip joint contact forces during normal gait ranging from 3.5 BW to approximately 8 BW (6,8,13,17,18,20).

In one study of patients that included total hip recipients, Brand and Crowninshield (2) concluded from their analyses that there was no statistically significant difference in the magnitudes of the joint contact force for normal subjects and postoperative total hip replacement patients 1 year after surgery. Furthermore, they found that there was no statistically significant difference between the joint contact forces for those two groups and the preoperative patients who were able to ambulate without aid. Their results predicted an average joint contact force of 3.6 BW for all three groups, 3.4 BW for preoperative, 3.7 BW for postoperative, and 3.5 BW for normal subjects with testing conducted at .4, .6, and .8 m/second, respectively. In another study, Paul and McGrouther (19) predicted peak joint contact forces for three patients at 6 months

FIG. 8. Force history for gait for patient 2 at 35 days post-surgery. Subject walked between parallel bars while sliding hands along bars. Two-legged support was present for 19% of the stride cycle, and the stride length was 1.22 m.



postsurgery. They reported force magnitudes of 2.7, 3.7, and 4.0 BW, but these values were not referenced to any walking speed.

Our gait data were collected during the early postoperative period from two patients walking at widely differing speeds ranging from a low of .5 m/second to a high of 1.8 m/second. The normal speeds of patients 1 and 2 were approximately .5 and 1.1 m/second, respectively, and most of our data were collected at these speeds. Despite the substantial difference in the speeds of our patients, we found that their joint loads were nearly identical, and both quickly increasing in magnitude during the first 3 weeks after surgery and then leveling off at

roughly 2.6 BW (Fig. 9). It was not until the speed of patient 2 exceeded 1.3 m/second that a pronounced speed effect became apparent. At 1.3–1.4 m/second, the average peak joint loads rose to approximately 2.8 BW, and at 1.8 m/second, the joint contact loads rose to 3.6 BW. This lack of a strong

TABLE 5. Average peak forces for rising from a chair

Patient	Days postsurgery	Type of support	Resultant force (\times BW)		
			Average ^a	Steps	Range ^b
1	31	Arms	1.2	2	
2	14	N/A	N/A		
2	23	Arms	1.20	2	0.80–1.60
2	35	Arms	1.16	2	0.94–1.38
2	44	Arms	0.88	2	0.80–0.96
2	58	Free	1.23	2	1.23–1.23

Chair height was 44.5 cm. Heels were roughly 6 inches apart with the toes pointed slightly outward. Weight was evenly distributed between both feet as judged by patients.

^a Peak force attained on each step for total steps.

^b Range is of peak forces over total steps.

TABLE 6. Miscellaneous activities

Activity	Patient	Days postsurgery	Force (BW)
Flexion (20–30°)	1	31	1.7
Extension (20–30°)	1	31	1.9
	2	58	1.8
Ipsilateral abduction (standing <30°)	1	31	1.2
	2	23	1.8
		58	1.9
Ipsilateral abduction (horizontal <30°)	2	58	1.2
Ipsilateral supine leg lift (low <30°)	2	58	1.4
Ipsilateral supine leg lift (high >45°)	1	31	1.8
	2	35	2.0
	2	58	1.5
Contralateral supine leg lift (high >45°)	2	35	2.6
	2	58	1.9
Double leg stance	1	31	0.8
	2	23	0.8

Horizontal ipsilateral abduction was performed with the patient lying on his contralateral side and raising the ipsilateral leg to an angle of approximately 30°.

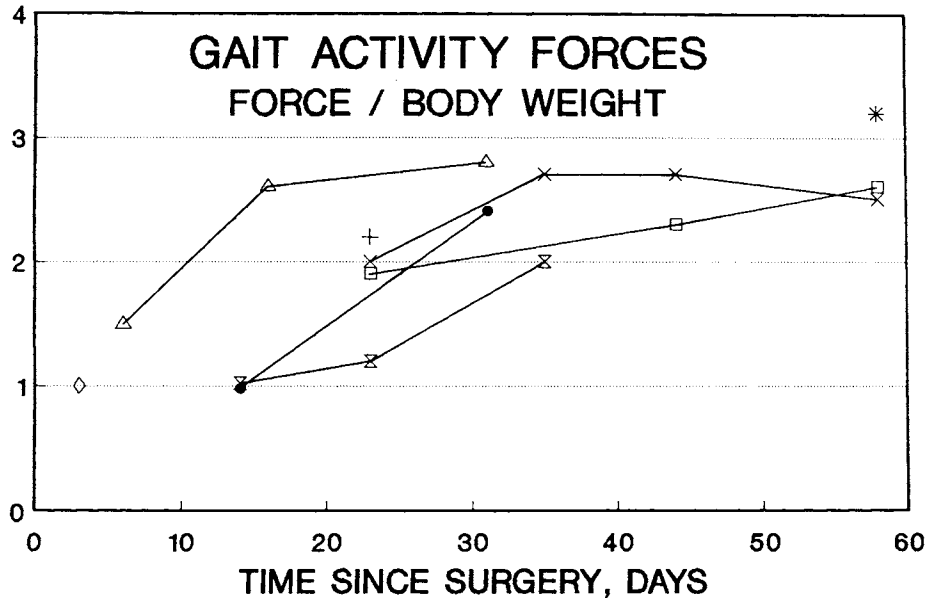


FIG. 9. —●—, 1-Par'l. Bars; +, 2-Par'l. Bars; *, 2-Fast Gait; □, 2-Cane; ×, 2-Free; ◇, 1-Walker; △, 1-Crutches; —×—, 2-Crutches. Peak loads during gait as a function of time post-surgery.

speed effect is in agreement with the results and conclusions of a study by Crowninshield et al. (7). In that study the predicted joint contact loads were much higher than we measured, but they found only a 15% increase in joint contact force when velocity increased from .9 to 1.4 m/second. The data from our second patient showed an increase in joint contact force of 17% for the same increase in velocity.

The walking speeds of our second patient were nearly equal to those of the two patients from whom Rydell obtained in vivo data at 6 months postsurgery. At our second patient's normal pace his joint contact forces fell between those measured by Rydell: 2.6 BW versus 1.6 and 3.0 BW. At patient 2's quick pace of 1.3–1.4 m/second, we measured an average peak joint contact force of 2.8 BW, which again was bracketed by the average values (1.8 and 3.3 BW) measured by Rydell. Taken in conjunction with the data reported by Rydell, our data suggest that in the range of walking speeds that are normal for a given individual, the peak joint contact forces are not strongly related to speed. Rydell found joint contact forces (referenced to BW) that differed by a factor of two for essentially identical walking speeds; and we have found joint forces that, at the individuals' normal pace, were virtually identical for walking speeds that differed by a factor of two. For our patients, this difference cannot be significantly reduced by normalizing the gait speed by some anatomical distance such as trochanter height or overall subject height.

Another activity for which there are in vivo as well as comparative analytical data is stair ascent. Rydell reported measuring peak contact loads that averaged 1.5 and 3.4 BW for his two patients. Only our second patient was able to ascend stairs without the use of bannisters for balance, and he generated peak contact forces that averaged 2.6 BW for eight cycles during the session at 58 days postsurgery. During that same session, for comparative purposes, he ascended the stairs using the bannister in his normal manner. During that activity the average contact force was reduced to 2.5 BW. The in vivo data from our patients and Rydell's patients are in close agreement, and all have produced data well below the magnitudes predicted by the analytic studies. In the study by Crowninshield et al. (8), a single patient was examined during stair ascent and the authors predicted joint contact force peaks that ranged from six to eight BW. They reconciled the discrepancies between their predictions and the in vivo data of Rydell (21) by arguing that it would be reasonable to expect the normal pain-free hip to accept loads of greater magnitude than the diseased or repaired hip. This argument does not appear to be supported by the later study of Brand and Crowninshield (2).

During ipsilateral single leg stance, our second patient showed some transient force peaks above five BW, but his average values for single leg stance were in the range of two to three BW. These were similar to the forces generated by our first patient,

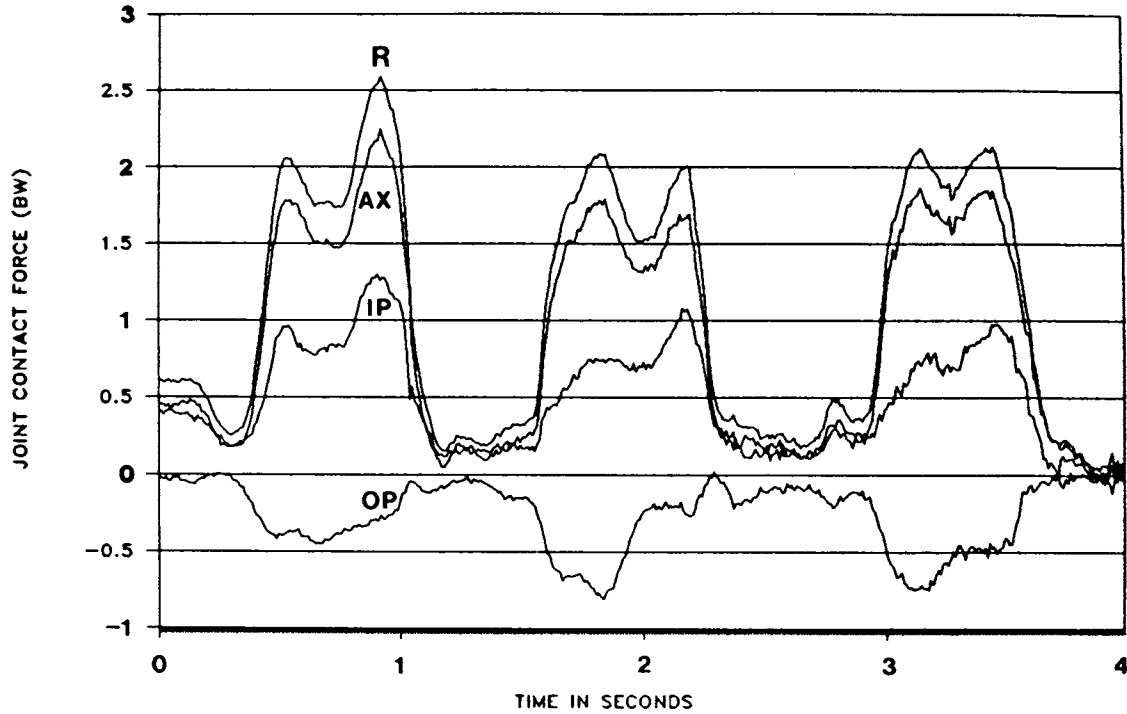


FIG. 10. Force history for stair ascent (three steps up) for patient 2 at 58 days postsurgery. Subject ascended step over step without use of bannister.

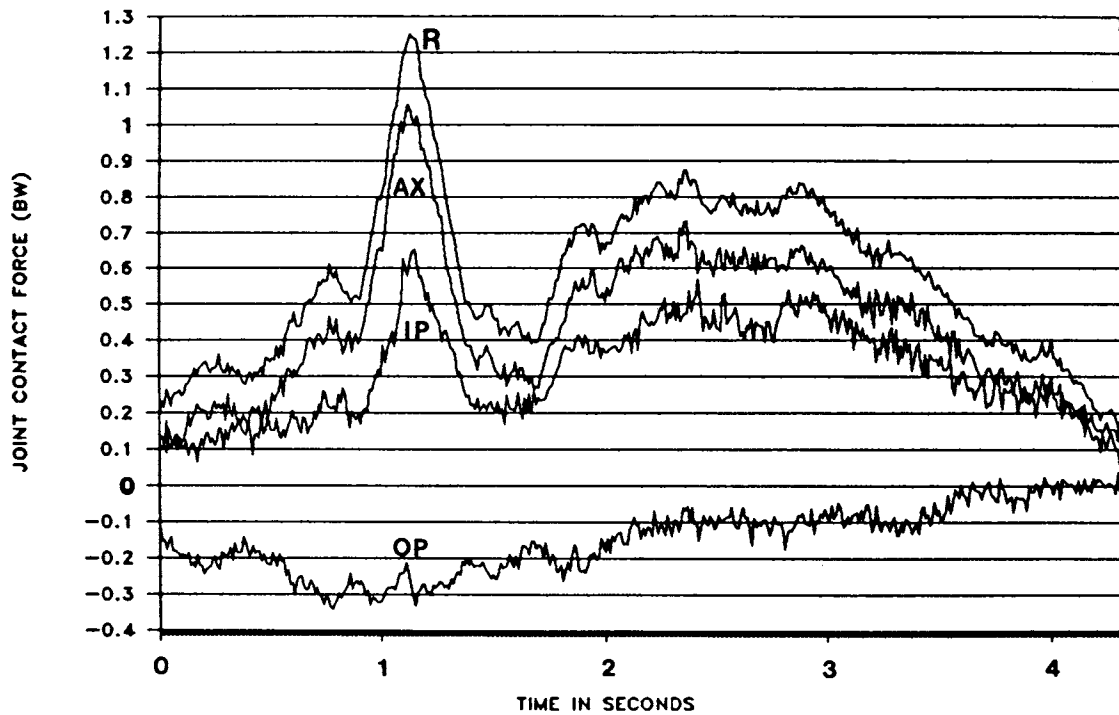


FIG. 11. Force history for rising from 44.5-cm high chair for patient 2 at 58 days postsurgery. Subject did not use any support in rising.

as shown in Table 2. We believe these magnitudes are typical of the forces required if unstable balance is not an issue. The analytic models have been static models and have not considered stability. Forces ranging as high as six BW have been predicted by models (15,22) for stationary single leg stance. Among the model studies, McLeish and Charnley (15) predicted force values most consistent with our results.

The results of the data for the two patients reported on here do not necessarily refute the results from the model studies because strategies for movement may vary substantially between normal subjects and those with total hip replacements (1). Our results do provide evidence that individuals perform these tasks during the early postoperative period while generating forces that are well below those typically predicted. A recent study by Brand et al. (3) used their gait analysis techniques to predict joint contact forces for our second patient. It was found in this study that certain modeling assumptions could be revised to bring predicted forces into closer agreement with the measured forces reported here. This work strongly suggests the need for additional experimental validation of the force prediction models.

The results of our study also suggest that rather than the activities most often considered, it is spontaneous or less practiced activities or those requiring balancing forces that can generate high hip joint loads in the postoperative subject. We reported previously that our first patient generated the highest loads during spontaneous maneuvers (preparing to move and reaching for her crutches) rather than planned motion sequences (11). For patient 2, we found that the highest force amplitudes (approaching 5.5 BW) occurred as he attempted to maintain his balance during single leg stance.

Finally, it is of interest to consider the characteristic bending and torsional moments generated by the measured contact forces. For the prosthesis used in this study, the moment arm about the base of the neck of the prosthesis for the two off-axis forces IP and OP is 38 mm. The characteristic bending moment for the vector sum of the IP and OP components for normal gait was 29.5 newton-meter (N-m) for patient 1 and 30.2 N-m for patient 2. The torsional moment arm about the stem for the OP force component is approximately 45 mm. The highest corresponding torsional moments that were generated were 15.4 N-m during normal gait for patient 1 and 21.4 N-m for patient 2 for normal gait.

The highest stem torques generated during stair ascent were 22.0 N-m for patient 1 (10,12) and 20.7 N-m for patient 2. Thus, whereas stair climbing induced higher stem torques for patient 1 when compared with level walking, patient 2 induced similar torques for stair ascent and level walking.

A word of caution must be inserted here regarding generalization of these moment and torque values. We believe they represent appropriate ranges for the early postoperative period, a period that is of considerable concern for early stability of non-cemented prostheses. Because the forces predicted by gait studies are higher than our measured forces, corresponding predicted moments and torques would be higher. The extent to which the differences in the joint forces and related quantities represent true discrepancies between in vivo data and models, or are due to the limited postoperative measurement times, will need to be determined from further study.

Acknowledgment: This work was supported in part by Grants AR35886 and AM00812 (RCDA) from the National Institutes of Health.

REFERENCES

1. Andriacchi TP, Galante JO, Draganich LF: Relationship between knee extensor mechanics and function following TKR. In: *The Knee, Papers of the 1st Scientific Meeting of the Knee Society*, ed by LD Dorr, Baltimore: University Park Press, 1985:83.
2. Brand RA, Crowninshield RD: The effect of cane use on hip contact force. *Clin Orthop Rel Res* 147:131, 1980
3. Brand RA, Pedersen DR, Davy DT, Kotzar GM, Heiple KG, Goldberg VM: Comparison of hip force calculations and measurements in the same patient, April 3, 1989. (unpublished manuscript)
4. Brown RH, Burstein AH, Frankel VH: Telemetering *in vivo* loads from nail plate implants. *J Biomech* 17:907, 1982
5. Brown RH, Kotzar GM, Davy DT, et al.: Three axis telemeterized prosthesis for *in vivo* force measurements across the hip. (*in press*)
6. Crowninshield RD: Use of optimization techniques to predict muscle forces. *J Biomech Eng* 100:88, 1978
7. Crowninshield RD, Brand RA, Johnston RC: The effects of walking velocity and age on hip kinematics and kinetics. *Clin Orthop Rel Res* 132:140, 1978
8. Crowninshield RD, Johnston RC, Andrews JG, Brand RA: A biomechanical investigation of the human hip. *J Biomech* 11:75, 1978
9. Crowninshield RD, Tolbert JR: Cement strain measures surrounding loose and well fixed femoral component stem. *J Biomed Mater Res* 17:819-828, 1983
10. Davy DT, Kotzar GM, Brown RH, et al.: Telemetric force measurements across the hip after total arthroplasty. *J Bone Joint Surg* 70-A:45, 1988
11. English TA, Kilvington M: *In vivo* records of hip loads using a femoral implant with telemetric output (a preliminary report). *J Biomed Eng* 1:115, 1979

12. Goldberg VM, Davy DT, Kotzar GM, et al.: *In vivo* hip forces. In: *Non-cemented Total Hip Arthroplasty*, ed by R Fitzgerald, Jr, New York: Raven Press, 1988: 251.
13. Hardt DE: Determining muscle forces in the leg during normal human walking—an application and evaluation of optimization methods. *J Biomech Eng* 100:72, 1978
14. Hodge WA, Fijan RS, Carlson KL, Burgess RG, Harris WH, Mann RW: Contact pressures in the human hip joint measured *in vivo*. *Proc Natl Acad Sci USA* 83:2879, 1986
15. McLeish RD, Charnley J: Abduction forces in the one-legged stance. *J Biomech Eng* 3:191, 1970
16. Morrison JB: The forces transmitted by the human knee joint during activity. Ph.D. Thesis, University of Strathclyde, Scotland, 1967
17. Paul JP: Approaches to design; force actions transmitted by joints in the human body. *Proc R Soc Lond [Biol]* 192:163, 1976
18. Paul JP: Forces transmitted by joints in the human body. *Proc Inst Mech Eng* 181:8, 1967
19. Paul JP, McGrouther DA: Forces transmitted at the hip and knee joint of normal and disabled persons during a range of activities. *Acta Orthop Belg* 41(Suppl 1):78, 1975
20. Rohrlé H, Scholten R, Sigolotto C, Sollbach W, Kellner H: Joint forces in the human pelvis-leg skeleton during walking. *J Biomech Eng* 17:409, 1984
21. Rydell NW: Forces acting on the femoral head-prosthesis: a study on strain gage supplied prostheses in living persons. *Acta Orthop Scand [Suppl]* 88: 1966
22. Williams JF, Svensson NL: A force analysis of the hip joint. *Biomed Eng* 3:365, 1968