In vivo hip joint forces recorded on a strain gauged 'English' prosthesis using an implanted transmitter

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This manuscript is based on the work and writings of the late Mr T. A. English of the Orthopaedic Department, Hull Royal Infirmary, Hull

This paper describes the results obtained from implanting a strain gauged version of an 'English' hip joint replacement together with a totally implantable FM radio transmitter. The implant is based upon a new concept in the design of femoral hip components having a diminished head offset to reduce head load and improved stem shape permitting alignment of the neck along the theoretical axis of peak load transmitted during the gait cycle.

The implant was inserted using the 'English' trochanteric approach (English, 1975) which further reduces the load on a prosthetic hip joint with the use of a spacer out from the redundant femoral head to rearrange the trochanteric muscle lever arms.

The resulting axial load is detected by four strain gauges mounted on a 'piston in cylinder' arrangement contained within the thickened neck of the prosthesis.

The single channel FM transmitter relays the gauge output to a signal processing unit to give a direct output of activity for recording on a UV recorder. Recordings were taken during implantation, recovery, walking (at three days) physiotherapy, stair climbing and walking over a period of forty days.

Introduction

The opportunity to carry out this type of load measurement arose during the development and successful clinical trials of the English prosthesis. The aim had been to develop a 'heavy duty' prosthesis having maximum fatigue resistance that would be suitable for the majority of patients especially the younger, active/heavy ones.

The prosthesis, when used with the English trochanteric approach (English, 1975), offers a significant improvement in prosthesis life by rearranging the trochanteric muscle lever arms to reduce the load and therefore the wear of the HDP (high density polyethylene) accetabular socket and by reducing the offset of the prosthesis ball to eliminate the large bending moments applied to a conventional 45 mm offset implant (Fig. 1). The femoral component stem profile was determined with the aid of photo-elastic models to indicate the stem stress distribution. A large neck collar is incorporated to allow the calcar to assist in transmitting the joint load to the bone.

The cross-sectional area of the implant is as large as possible in order to resist the complex backward bending and torsional forces present during normal activity.

Stress raisers in both the bone cement and the implant have been minimized by radiusing the stem corners and tapering the stem on all sides to cause the cement to be compressed during insertion. A standard 37 mm prosthesis was modified to include a 'piston in cylinder' component. The piston is fitted with strain gauges to determine the forces acting upon it during various activities. The output of the strain gauge bridge is relayed by a miniature battery powered transmitter to a FM receiver over a maxamum range of nine metres. The output of the receiver is connected to a UV or pen recorder giving a direct print-out of load patterns.

Background

Rydell (1966) carried out recorded work of a similar nature in two patients using a modified large headed (47.2 mm), non-cemented Austin Moore type partial hip replacement having strain gauges mounted inside a lengthened neck (Fig. 2). The resulting strain together with limb angle ciné film recordings and force platform data was used to calculate the resultant hip joint forces during walking. Measurements started six months after operation and were limited to one week by the need to bring the signal wires out through the skin for processing. The range of dynamic activities studied was limited by the length of the electronic walkway. Paul (1966, 1971) analysed dynamic loads during walking using a force platform and ciné film records of the joint angles. Further work by Paul and McGrouther (1976) showed how the peak load in walking occurred just before toe off or at 47 per cent of cycle time.

Carlson, Mann and Harris (1974) described an *in vivo* system for recording cartilage surface pressure. The implant selected (an Austin Moore type) had fourteen pressure transducers mounted inside the hollow head of the prostheses. As yet, no *in vivo* studies using this system have been described.

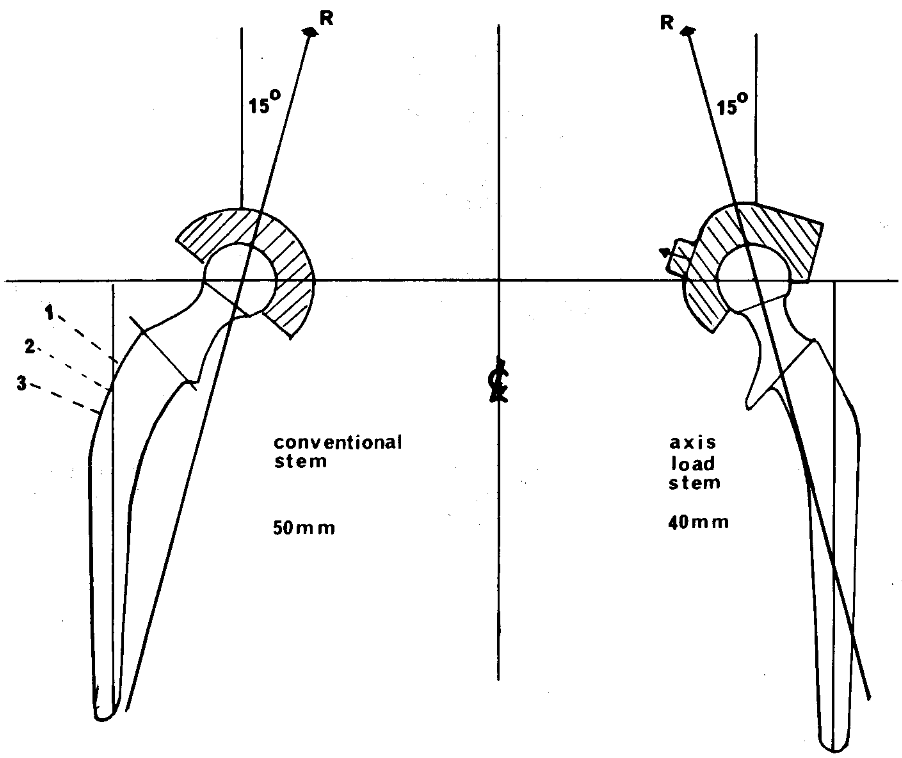

Fig. 1. Resultant load directions axial and conventional prosthesis

Need for further hip load investigations

The need for further investigations into the forces in the hip prosthesis arose during the development of the 'English' implant (1976). Increasing numbers of fractured Charnley first generation stems were also being observed. Examination of the fracture stems in our possession by metallurgists at Messrs Hawker Siddeley, Brough and AWRE (Atomic Weapons Research Establishment) (Jordan, 1976, 1977) indicated that the stems had failed by a reverse bending type of fatigue. (Fig. 3). This is consistent with the very low yield strength, in relation to its ultimate tensile strength, of 316L stainless steel. We tried to reproduce similar fractures using a specially built hip joint loading machine. This applied an average load of 2891 N at rates of 230–240 cycles per minute for 10 000 000 cycles to a standard first generation flatback Charnley prosthesis, the stem of which had been imbedded to a depth of 82 mm in concrete. The long axis of the stem was placed vertically and the load applied to the 47 mm offset head which deflected elastically 0.7–0.8 mm during each load cycle. Even with initial gross overloading resulting in plastic deformation, it was impossible to produce a fatigue failure. Similar tests at different angles and loads also failed. The literature summarized by Walker (1974) gives walking forces varying between 1.8 and twelve times body weight. Barr *et al* estimated that 8 500 000 weight bearing cycles were placed upon an implant in five years and Scales and Lowe (1971) estimated that 1–2.5 million cycles occurred in one year.

Axial loading of the femur

The 'first peak' early in the cycle (7 per cent) after heel strike, suggested by static analysis and Paul's earlier work occurs with the hip loaded in zero degrees of flexion. Paul's later experiments showed that the higher 'second peak' occurs later in the cycle (47 per cent) at 70° of hip flexion before toe off (Fig. 4). In this flexed posture, despite femoral neck anteversion, the maximum load is transmitted axially approximately down the line of the calcar femoral in all three axes of pelvic loading. More detailed muscle force analysis with special reference to trochanteric repositioning is discussed elsewhere. (English, Dowson, 1981; English, Dowson, Jobbins, 1975).

Method

An artificial hip joint was designed so that its neck portion would lie along the axis of the theoretical maximum dynamic load. The maximum axial load is intended to be compressive and is measured by a sealed piston component fitted with four strain gauges, (Fig. 5), the output of which is relayed by a single channel transmitter making a unit with maximum protection against corrosion and fatigue after implantation (English *et al*, 1980; English *et al* 1979; Goodman *et al*, 1979). To reduce the neck-shaft angle in the transverse (coronal) plane, the implant required an offset less than any previous clinical model of prosthesis. Photo-elastic studies (Sharples, Kilvington and English, 1976) were used in the stem design analysis. These tests showed that the theoretical axial loading produced pure compression on the neck portion, (Fig. 6). Further work using epoxy-resin models showed that the maximum tensile stress further down the stem portion (i.e. 80 mm from the top of the 22 mm head) could be reduced from 76532 to 47574 N/m^2 × 10 when compared with the original Charnley 'flat back' design, the load being applied to the top of the head, and parallel to the stem. These tests gave

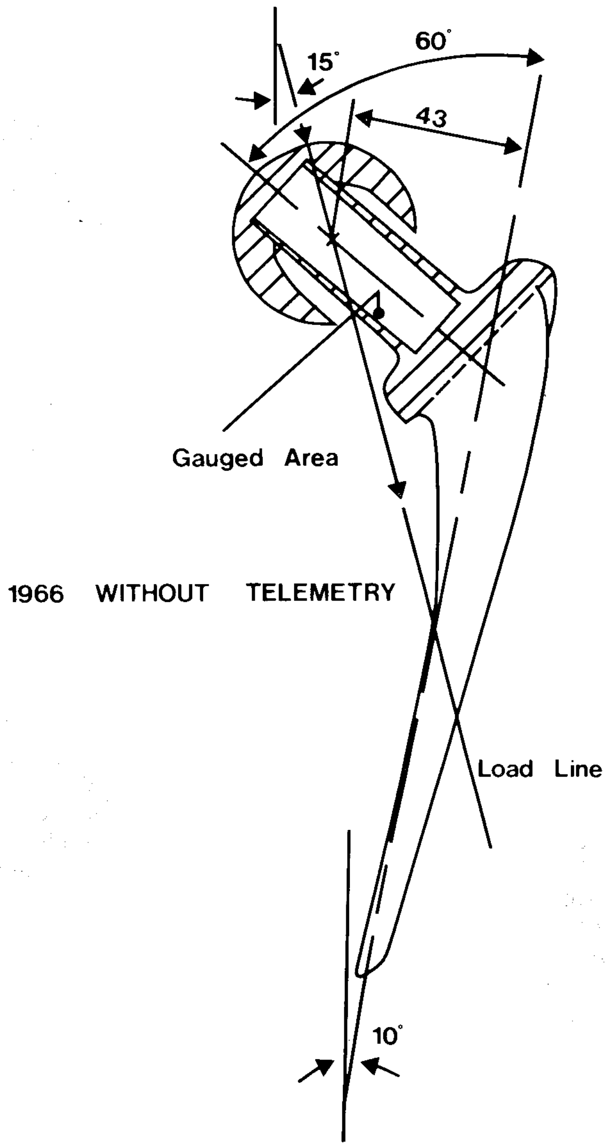

Fig. 2. Rydell strain gauged prosthesis (1966)

a calculated improvement in stem strength of 38 per cent and led to the development of the current clinical design in which there is evenly distributed compressive stress down to the tip of the stem at theoretical maximal loading postures.

Transmitter design

Strain gauges

The main factor governing the type of transmission scheme used, is the type of strain gauge employed. The 'English' prosthesis is manufactured in EN316 stainless steel which yields at a strain level of approximately 0.04 per cent. At a force of 2224 N the strain expected is 0.0112 per cent. This means that the maximum possible change in gauge resistance ΔR, relative to original resistance R, is given by: $\Delta R/R = eK$ where e is the strain and K is the 'gauge' factor which lies between 2 and 6 for metal gauges. Thus, for example, with a high output platinum–tungsten gauge which has $K = 4.5$, the maximum ratio $\Delta R/R$ is 0.2 per cent. In addition, this value is the maximum $\Delta R/R$, and in normal operation the loads

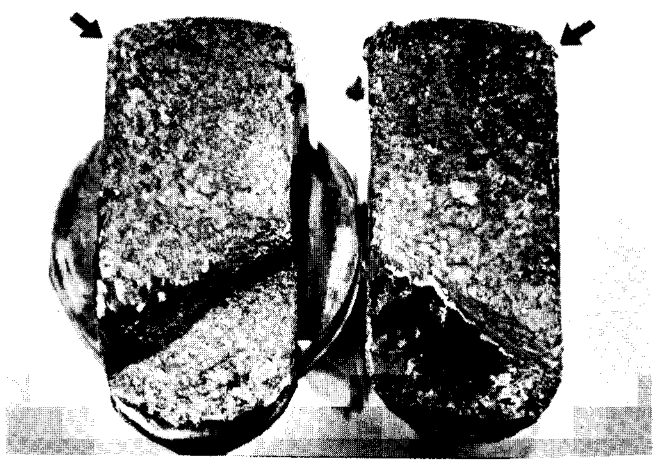

Fig. 3a. Fractured Charnley type prosthesis showing origin of fracture initiation point

Fig. 3b. Metallographic structure of fractured prosthesis

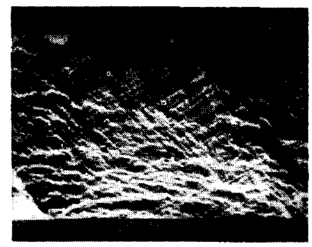

Fig. 3c. Fatigue striations on fracture surface (B. A. Jordon, AWRE 1976)

expected would cause a change in resistance considerably less than this (say 100 times less) it is therefore necessary to operate several strain gauges in a bridge arrangement in order to produce a suitable output signal.

Figure 7 illustrates the solution adopted and shows the prosthesis which is different from the normal clinical model in that it is manufactured in two parts. The head part forms a removable piston and fits into a cylinder machined out of the specially thickened neck of the femoral component. The removable neck is machined to allow for four miniature Dentronic 1800 series platinum tungsten strain gauges ($R = 350$) to be mounted around its circumference thus forming a simple load cell. The gauges are placed 90° apart around the circumference of the load cell. Each one has its gauge length axis at right angles to its neighbour. In axial compression all four gauges detect strain, although the two having their gauge length axis aligned around the circumference do so to a lesser extent. During front/back directed side loading, the resulting combination of bending and compressive strain is picked up by these circumferential gauges. No temperature compensation is included in the bridge for reasons of simplicity, and because the body should remain at a sufficiently constant temperature to make compensation unnecessary for reasons of accuracy.

Modulation scheme

In this application the information to be transmitted is essentially dynamic d.c., i.e. loading of the prosthesis will produce a certain d.c. output from the gauge bridge.

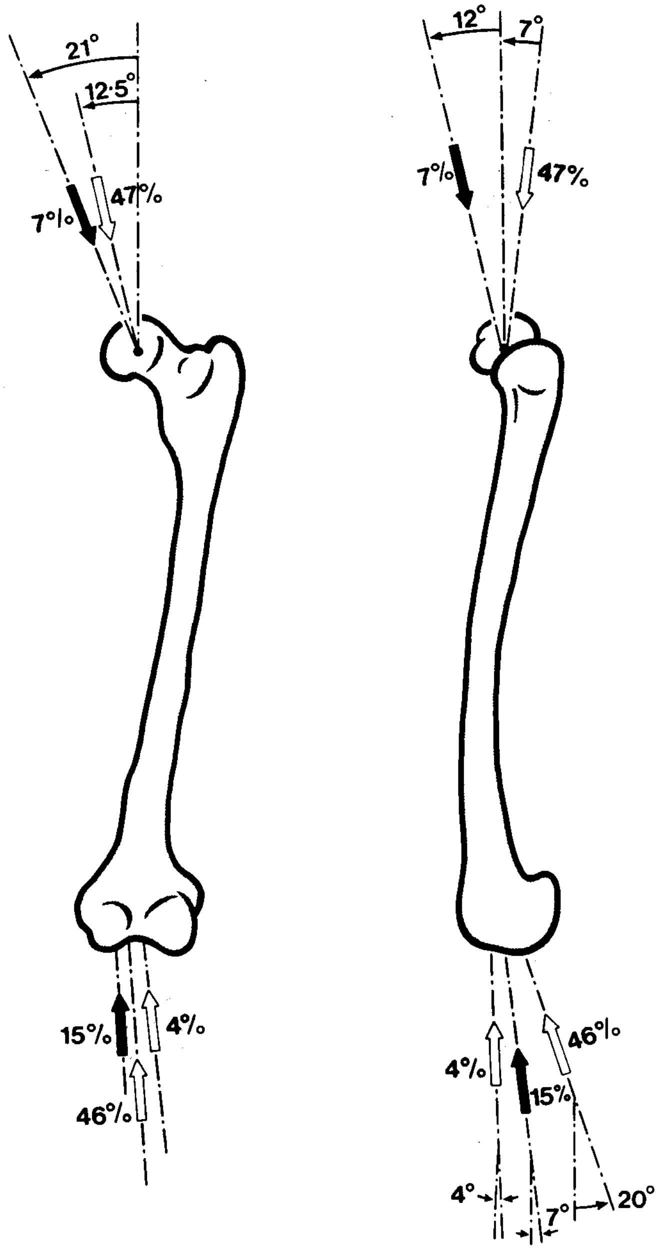

Fig. 4. Peak femur loading at 7° of hip flexion before toe off (J. P. Paul, 1976)

Following normal telemetric practice we should therefore use the bridge output to drive some form of pulse duration modulator, whose output pulse train can then modulate an FM transmitter. In this way information is in the mark/space ratio of the pulse train, and is independent of amplitude levels and hence battery life. This solution was not adopted for several reasons. The bridge output from no load to maximum is only about 0–3 mV per volt input. In order to drive a pulse duration modulator this signal would have to be amplified, thus requiring more transmitter circuitry and giving lower battery life. Alternatively a high bridge input voltage would have to be used, and this would require a physically larger transmitter. The final solution was to adopt a simple amplitude modulation system, and to check that accuracy was maintained over a reasonable proportion of the battery life by extensive pre-implantation testing.

Transmitter system

Figure 8 shows the final transmitter circuit design. The LM3909 oscillator chip is powered by a single 1.35 V battery (Duracell mercury WH 3T2) 220 mA hours, and produces a 1 V amplitude 1 kHz square wave which is used to excite the strain gauge bridge. The output of the bridge is balanced to about 0.5 mV peak-to-peak by means of the parallel balance resistor. The bridge output is approximately 3 mV at full load (9000 N), and is very noisy due to the inherent chip noise in the excitation signal. The bridge output is used to modulate directly an FM transmitter chip (Sandev SN102F) which operates in the VHF bio-telemetry band at 102.3 MHz. The maximum input amplitude is 4 mV, so that bridge and transmitter are well matched. Both the oscillator (1.5 V) and transmitter (2 × 1.35 V) battery circuits are fitted with sub-miniature reed switches which enable power to be switched on and off by a magnet at distances of between 25–50 mm from the transmitter. Battery life is approximately seventy hours. Figure 9 shows the mechanical construction of the transmitter which is approximately 25 × 35 × 10 mm and Fig. 10 shows the implanted system. Encapsulation is of paramount importance in any human implant and the required medical standard is achieved as follows. The gauges are sealed from the body fluids by coating with a medical grade silicone adhesive. The coating is allowed to cure and is applied slightly proud of the piston diameter. A further seal is provided by the addition of a groove in the neck and piston components, which are machined to line up on assembly. These grooves are filled with medical grade silicone rubber and allowed to cure, again slightly proud of the piston diameter. The PVC wires attached to the strain gauges are covered by a thin walled silicone rubber sleeve which is sealed into the piston centre hole with silicone adhesive. The cable is passed through the

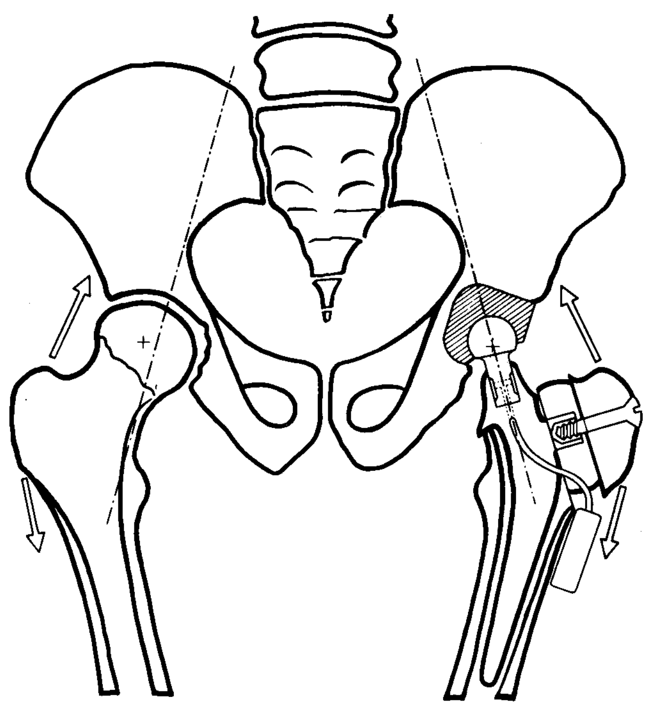

Fig. 5. Method of measuring the prosthesis axial force

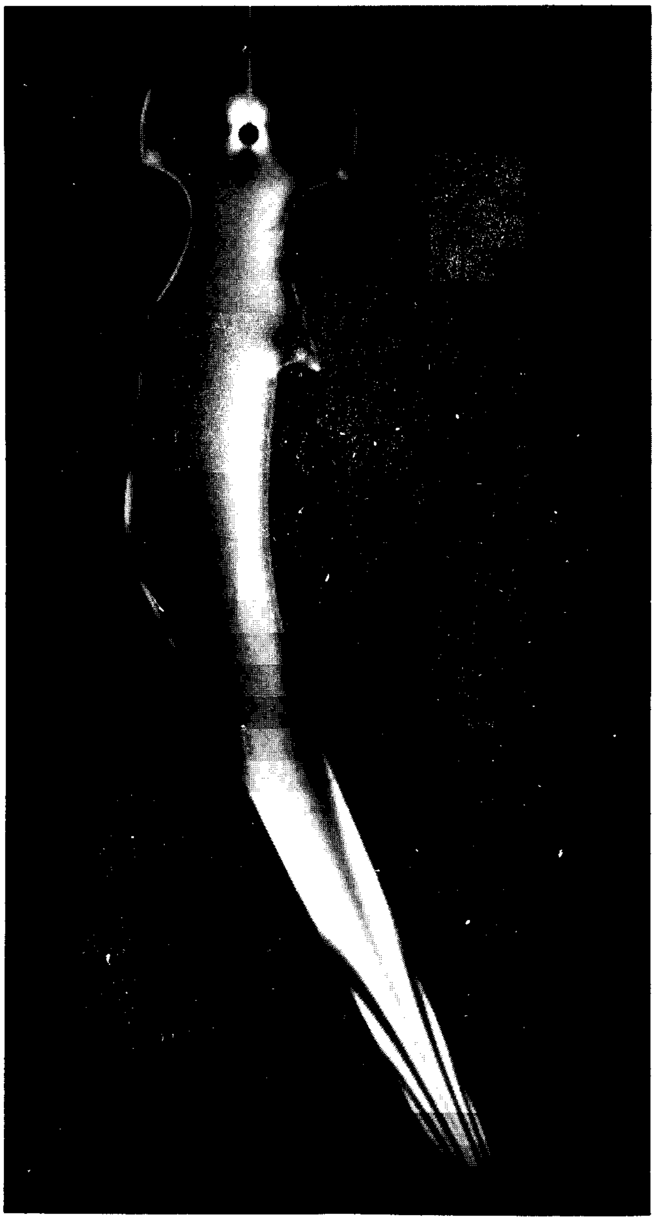

Fig. 6. Photo-elastic model of prosthesis under axial loading

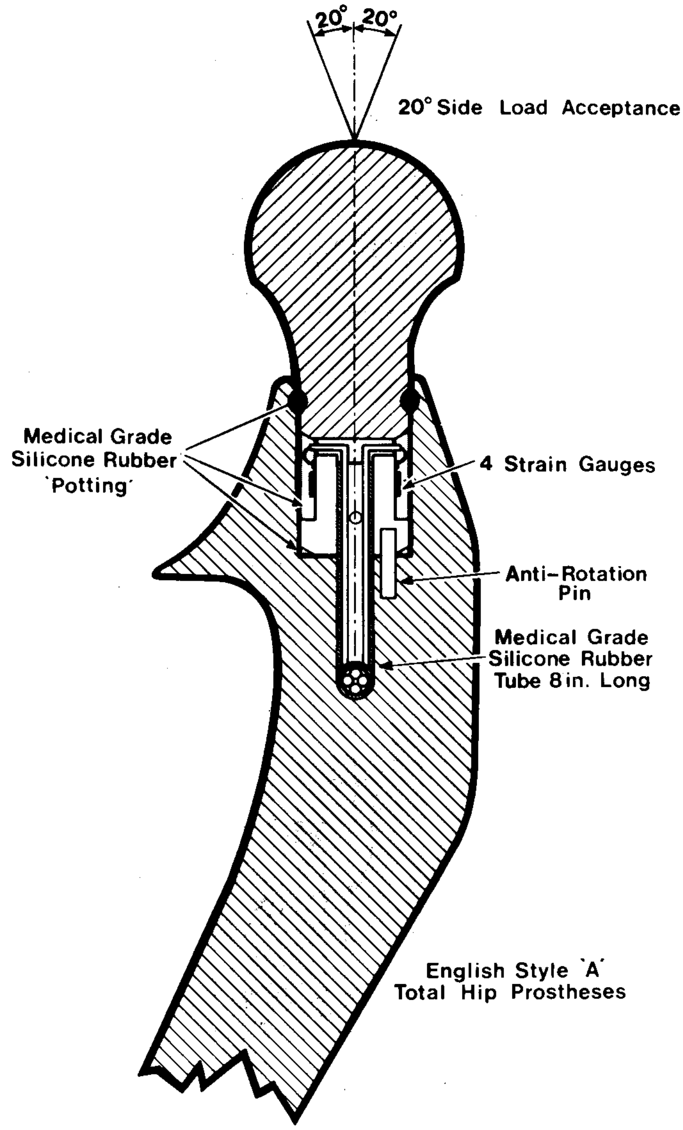

Fig. 7. Cross-sectional view of prosthesis head/neck

implant stem with the silicone sleeve protruding about half an inch. The main length of the PVC wires is then coated with silicone rubber adhesive and a thick walled silicone rubber tube passed over them to butt up to the implant stem and over the short length of thin walled tube. The signal cable is then attached to the transmitter and the perspex box is filled with epoxy-resin. The whole unit and cable are then covered in a silicone rubber.

The receiving system

Figure 11 shows the receiving system. The FM signal from the transmitter is received on a simple 3 in loop antenna taped to the patient's skin. The antenna cable is long enough to allow complete freedom of movement when the patient is undergoing tests. The antenna signal is received and demodulated using standard Mullard modules. The noisy square wave receiver output is then passed to the main filtering circuits and to an audio amplifier/loudspeaker system. The signal is processed in the following way to provide a noise-free d.c. signal whose amplitude varies in sympathy with the amplitude of the receiver output square wave. This d.c. signal is required to drive the chart recorder. The noisy square wave is first passed through a 1 kHz Q = 10 bandpass active filter to remove all high frequency noise. The resulting 1 kHz output is buffered and fed to a precision full wave rectifier. The rectifier output is therefore a d.c. signal with a large amount of 2 kHz ripple. This signal is sufficient to drive the tuning meter. The signal is then smoothed by the integrating action of the 15 Hz low pass filter. The output is now d.c. with low ripple. The next stage performs level shifting and attenuation to provide a signal suitable for driving the UV recorder.

Calibration and accuracy

The implant and telemetry system were extensively calibrated and tested before implantation. A modified fatigue testing machine was used for calibration. The machine was capable of applying known static or dynamic loads of up to 9000 N. Four ranges of load were recorded using a force applied axially down the implant

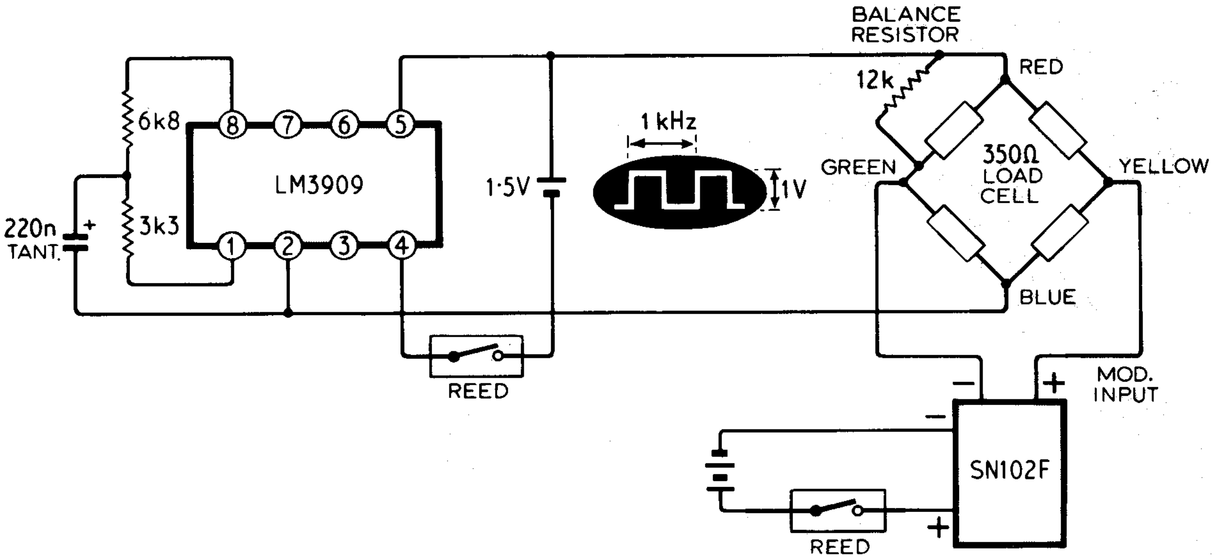

Fig. 8. Transmitter circuit diagram

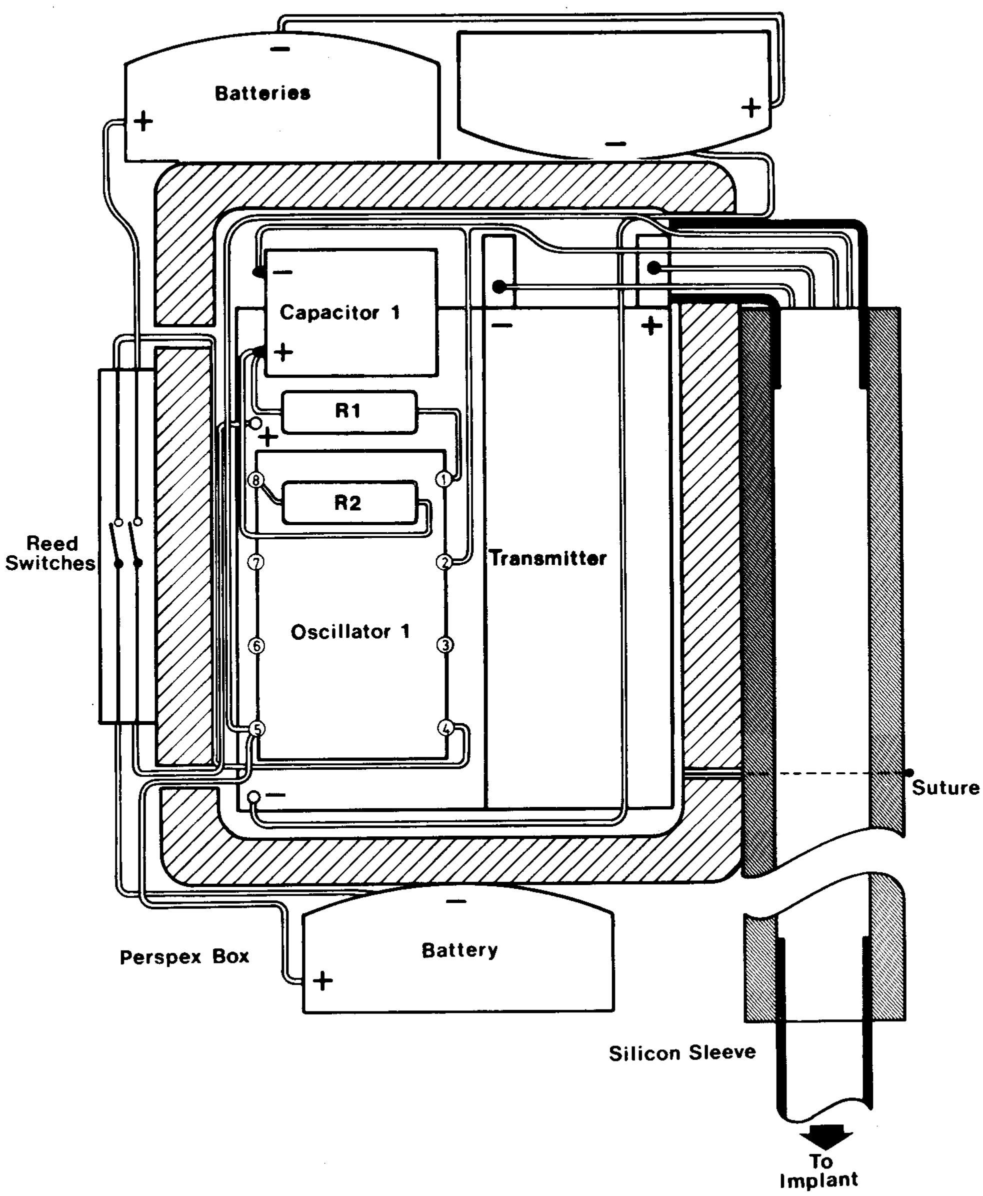

Fig. 9. Layout of implanted electronics package

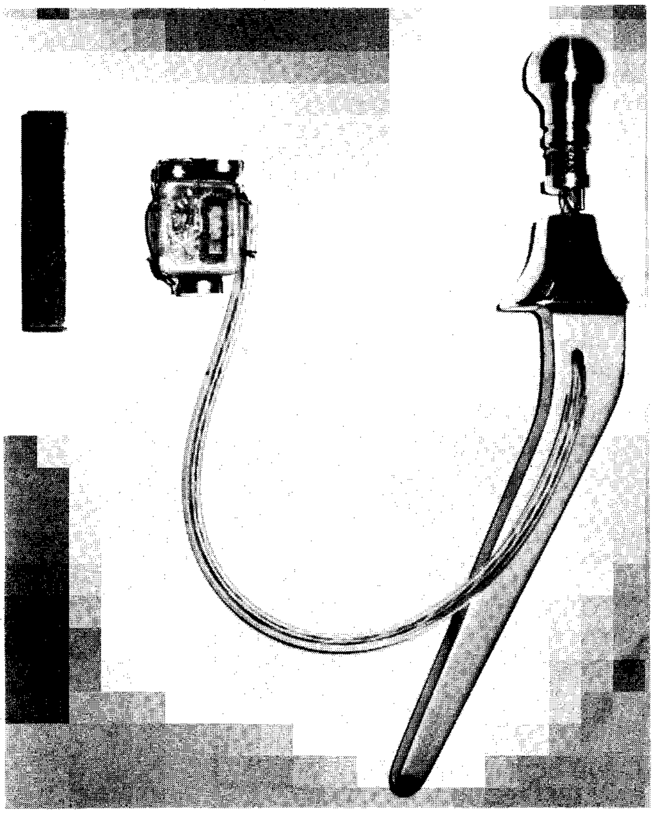

Fig. 10 Complete instrumented prosthesis before encapsulation of electronics and gauges

head and neck. The ranges were: 0–2224 N, 0–4448 N, 0–6672 N, 0–8896 N. In addition, loads were applied at 20° angles to the axis within the range 0–4448 N. This 20° off-axis loading was felt to be the greatest direction at which a force would be applied to the implant head during normal walking or high load activities, and gave rise to an apparent reduction in force of 7 per cent when compared with axial loading (Fig. 12). The system accuracy is affected by several factors. Firstly, the strain gauges are not temperature compensated, and the transmitter frequency varies with temperature. Secondly, because of the amplitude modulation system used, receiver tuning (and hence transmitter temperature) affects the output signal. Thirdly, off-axis forces cause an apparent reduction in load. Fourthly, metal deformation and gauge 'creep' will affect results in high force activities such as running. However, the system gave dependable and repeatable results when tested at body temperature (37°C), over several months, and over the period of transmitter battery life. An overall resolution of 40 N in the range 0–2224 N was obtained, and the overall system accuracy is estimated at plus or minus 10 per cent.

Implantation

The patient selected for the operation was female and weighed 79.8 kg. She had had one hip replaced three months previously using the English trochanteric approach. The nature of the operation and the likelihood of a revision to recover the telemetry package and to replace the gauged head of the implant with an ungauged one was fully explained to her. The calibrated implant was immersed in a solution of glutaraldehyde (Cidex) for two hours prior to operation, with the gauged neck exposed. The transmitter pack was sewn into a prepared autoclaved envelope of monofilament knitted polypropylene mesh (Marlex). Using the standard English trochanteric approach to the hip the femoral component was placed in 30° of anteversion and a standard English socket (22 mm diameter) inserted. A bone spacer 15 mm thick was cut from the redundant femoral head and used to space out the greater trochanter (Fig. 13). Operating time was two and a half hours during which an axilla temperature of 35.3°C was recorded.

Signal recording

Records were made during the operation using the attenuated output from the receiver to drive a high speed UV recorder or alternatively a high performance pen recorder. The UV recorder galvanometers used had

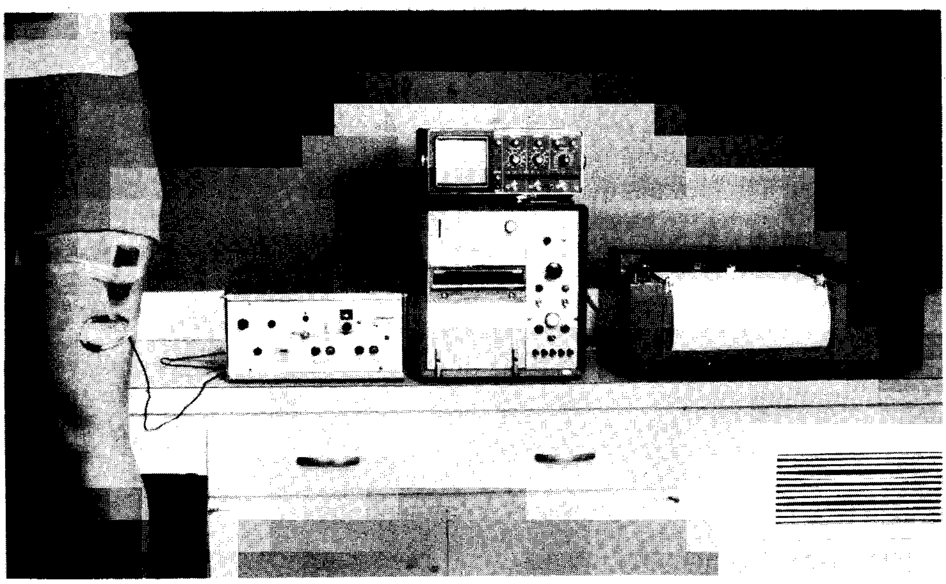

Fig. 11. The telemetry system in use

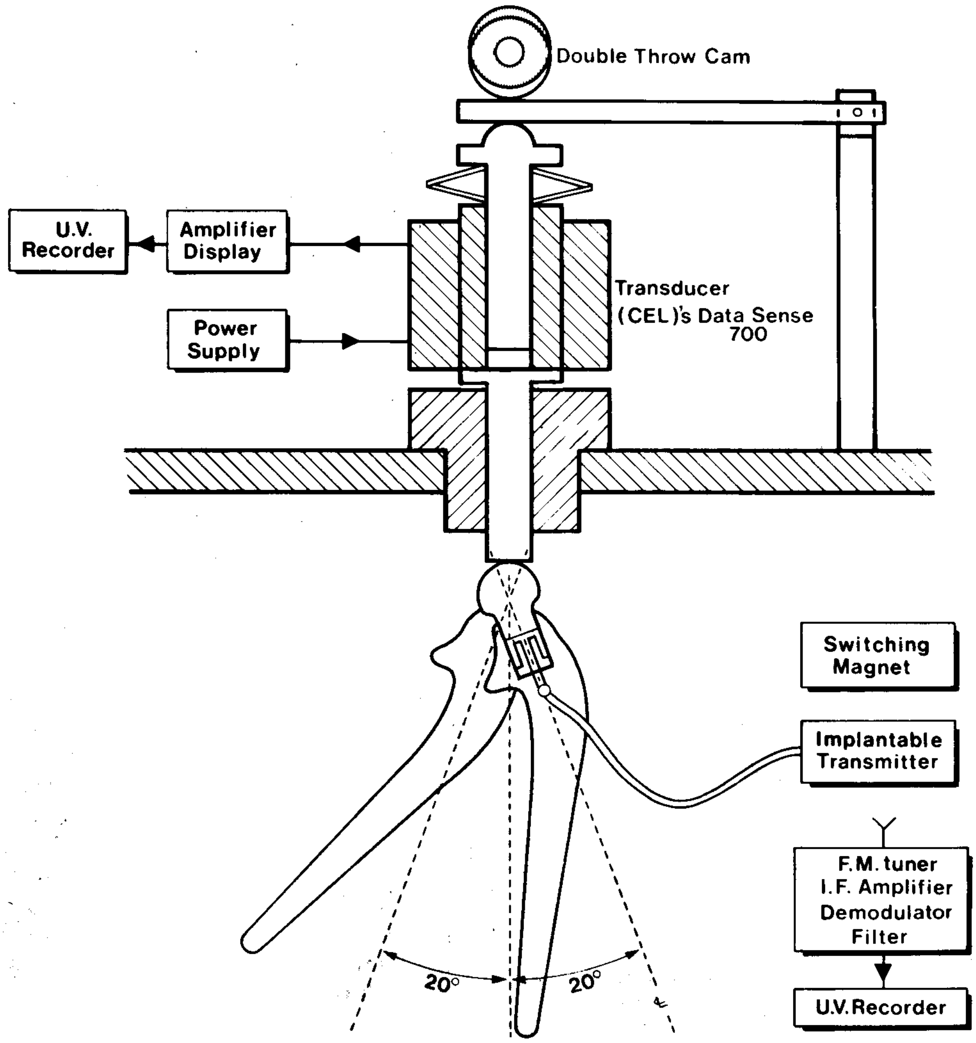

Fig. 12. Calibration method

a sensitivity of 0.050 mA/cm and a frequency response of 200 c/s. The transmitter was switched on using a sterile magnet placed within 25–50 mm of the telemetry pack. An aerial coil placed near the transmitter was used to obtain the best quality signals. Pre-operative testing had shown that the signal was 50 per cent lower at 37°C than at the ambient theatre temperature of 20°C. This effect was minimized during operation by burying the transmitter within the fat layer between test recordings.

Results

The base line signal for subsequent records was taken after allowing the transmitter to stabilize to body temperature in the fat and after cementing the components in place but with the hip still dislocated. The transmitter frequency changed if its temperature was lowered by removing it from the fat. After reduction at the stabilized temperature the load increased from zero to 0.2 times total body weight. Re-attaching the trochanter by bolting it to the implant and suturing the deep fascia over the trochanter gave a further small increase in force and there was thought to be further slight increase due to rise in the patient's body temperature. The increase due to the change in body temperature from 25°C to 37°C was allowed for in determining the base line for all subsequent records (Fig. 14a). Flexion of the hip in the supine posture on the operating table with the patient still anaesthetized caused a force of 0.3 times total body weight.

Day 0 to 3 hours post-op

Forces of about 0.3 times body weight were recorded during this period with the patient supine in bed and performing assisted flexion and rotation movements of the hip. Traction reduced the force to zero (Fig. 14b).

Day 1

Early in the first post-operative day the force had increased to 0.61 times body weight which was considered to be due to an increase in muscle force partly due to excretion of the muscle relaxant used at operation (pancuronium bromide). Movement of the hip supine in bed showed further increases in force (Fig. 14c).

Day 2

Recorded forces were similar to those taken on Day 1.

Day 3

Forces of 0.68 and 0.78 times body weight were

Fig. 13. Radiograph post operation

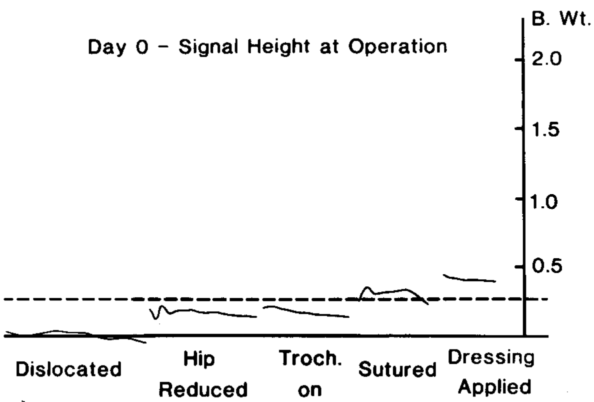

Fig. 14a. Load recording during implantation of prosthesis

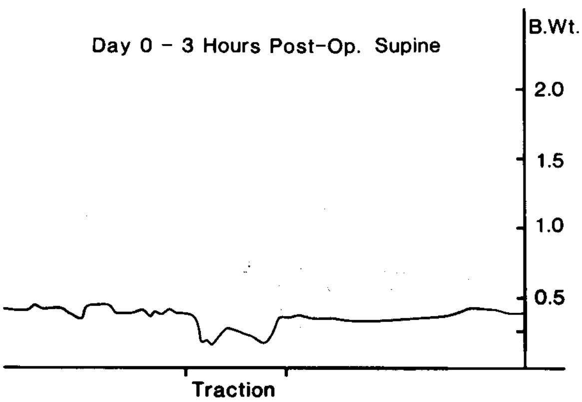

Fig. 14b. Load recording showing the effect of traction

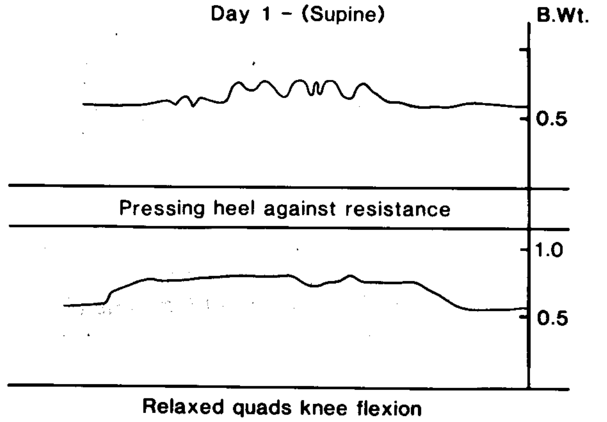

Fig. 14c. Load recording showing the effect of physiotherapy

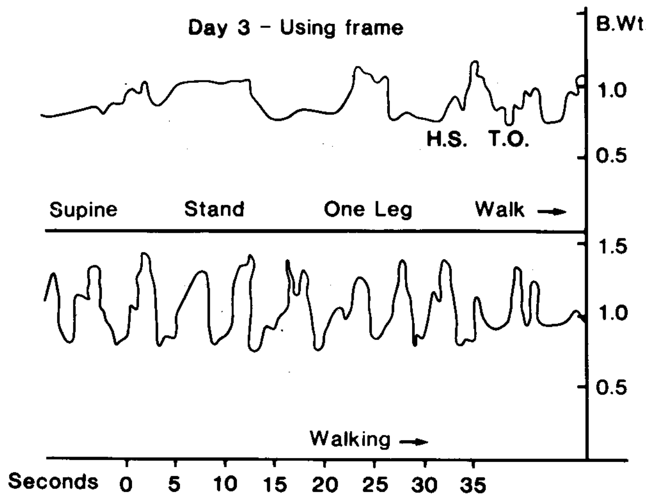

Fig. 14d. Load recording during first assisted walking attempts

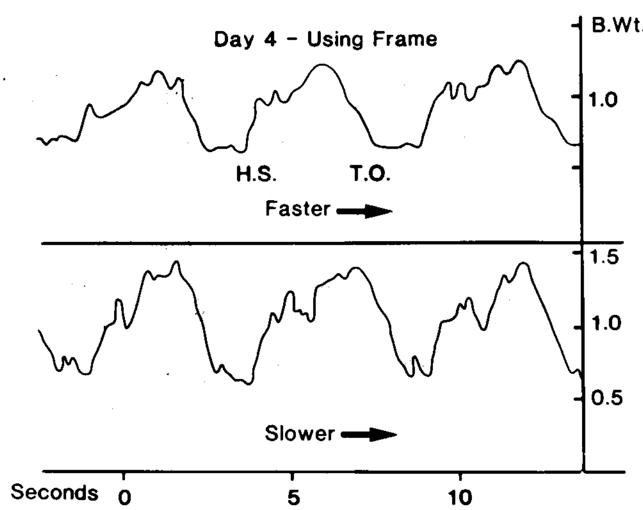

Fig. 14e. Load recording during assisted walking

recorded with the patient supine. Standing for the first time leaning on a walking frame produced a force of 1.16 times body weight. Walking forces on this day ranged from 1.35 times body weight during the stance phase to 0.7 times body weight during the swing phase (Fig. 14d). There was a further increase on attempting to stand on the operated leg with support.

Day 4

Supine resting force was 0.81 times body weight. Walking with a frame the forces during the stance phase were 1.4 times body weight and during the swing phase 0.7 times body weight (Fig. 14e).

Day 7

Walking faster using a frame the stance phase force was 1.55 times body weight and the swing phase 0.6 times body weight. The highest peak measured 1.8 times body weight.

Day 12

The supine force was 1.0 times body weight. Forces recorded during supine physiotherapy including static quadriceps, planter flexion and dorsiflexion and hip flexion to 30° and 60° are shown in Table 1. The one-legged stance showed a marked increase to 2.54 times body weight with support and 2.8 when attempting to stand briefly without support. This static one-legged stance force exceeded all the dynamic walking forces recorded. One-legged stance on the opposite leg resulted in a force of 1.44 times body weight. Stance forces with the heels separated 10–15 cm were 1.43 times body weight. Walking up 10 cm stairs gave stance forces of 1.92 times body weight and swing phase forces of 0.099 times body weight. Walking down the same stairs gave stance phase forces of 1.79 and swing phase forces of 0.99 times body weight (Fig. 15a). The patient was able to walk unaided but was more confident using one stick. There was little difference in peak force whether the stick was used or not (Fig. 15b). On this occasion stance phase forces averaged 2.0 times body weight and swing phase forces 0.93 times body weight. These forces were slightly higher than those recorded later. Sitting in a chair the force record was 0.98 times body weight. On the twelfth post operative day the patient was allowed home and returned to work as a Cashier.

Day 35

An unidentified form of interference affecting the transmitted signal was detected intermittently throughout this session. Clear records were made during static stance with heels 20 cm apart showing forces of 1.55

Table 1. Forces during physiotherapy (Force in terms of body weight)

Load condition		Days post operation 0	1	2	3	4	5	7	12	35	36	40
Assisted leg movements	High	0.3										
	Low	0.0										
Leg pressing resisted	High		0.9		0.9		1.2					
	Low		0.6		0.7		0.9					
Relaxed quads	High		0.8									
	Low		0.6									
Knee flexion	High		0.8	0.8								
	Low		0.8	0.6								
60°	High								1.3			
	Low								0.9			
30°	High								1.4			
	Low								0.9			
Static quads	High			0.8					1.2			
	Low			0.6					0.9			
Feet circling	High			0.7								
	Low			0.6								
Dorsi flexion	High								1.4			
	Low								0.9			
Planter flexion	High								1.1			
	Low								0.9			
Resisted inversion	High								1.4			
	Low								0.9			
Straight leg raising	High											1.3
	Low											1.0
Hip flexion	High											1.2
	Low											1.0

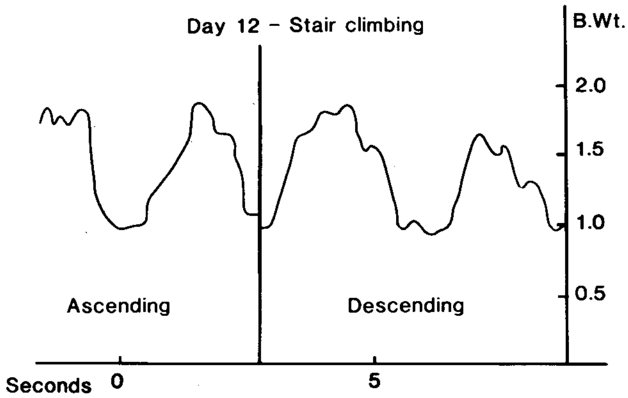

Fig. 15a. Load recording during stair climbing

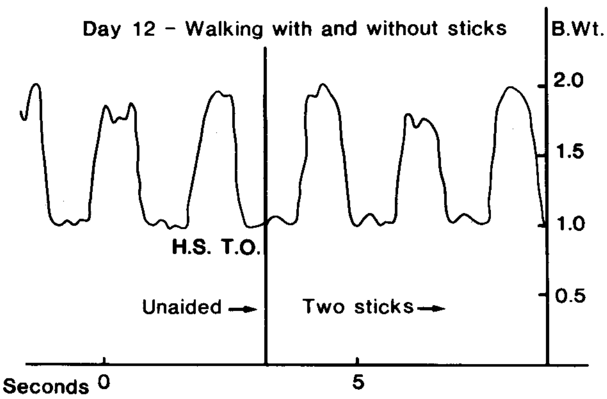

Fig. 15b. Load recording during walking with and without sticks

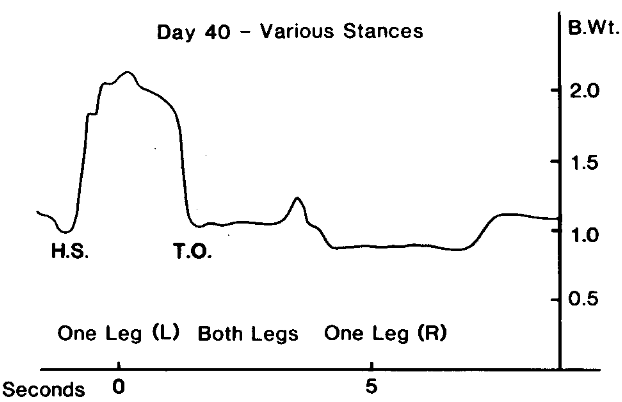

Fig. 15c. Load recording during various stances

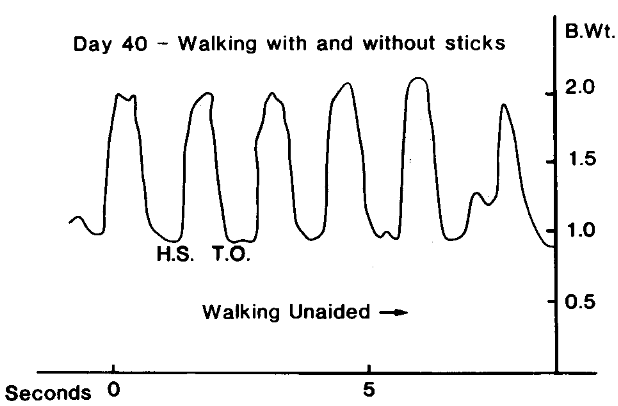

Fig. 15d. Load recording during walking with and without sticks

times body weight, stance phase forces of 1.8 times body weight and swing phase forces of 0.93 times body weight were recorded when walking. Difficulties were encountered in switching on the transmitter with the original magnet on this day. Radiographs showed that the transmitter pack had moved inwards approximately 12 mm. A change in the transmission frequency had also occurred although at this stage only eight of the expected seventy hours of battery life had been used.

Day 36

Standing forces were similar to those taken upon Day 35 and the stance and swing phase walking forces were similar. One-legged stance forces of 2.61 times body weight were recorded. Records were subject to the same interference as on Day 35.

Day 40

During this recording session there was no signal interference. Forces of 1.0 to 1.3 times body weight were recorded during supine straight leg raising and hip flexion. Static stance with heels 20 cm apart caused forces of 1.26 times body weight. One-legged stance on the operated leg gave a force of 2.18 times body weight, i.e. lower than on the previous occasions, possibly explained by better muscle control of pelvic tilt and posture thereby reducing the load (Fig. 15c). Comparison was made walking with and without sticks. Again no significant change in force was detected whether sticks were used or not. There was some minor indication of the two peak pattern of gait cycle described by Paul though it was not so marked on Day 40 as it had been on earlier occasions walking at lower speeds. The two peak effect was absent when walking without sticks (Fig. 15d). At this stage the stance phase represented 43 per cent of the cycle. The various forces recorded are summarized in Tables 1 and 2.

Conclusions

After Day 40 no further signals were received. It was felt unjustifiable to subject the patient to an operative procedure to retrieve the transmitter and the gauged portion of the implant at this stage since she had no adverse symptoms. A possible cause for failure may be in the type of medical grade silicone rubber used for encapsulating the strain gauges as it has since come to our notice that the acetic acid present in this grade of uncured silicone could have corroded the gauges or solder connections. A specialist epoxy strain gauge precoat followed by a non-corrosive silicone rubber compound (Dow Corning 3140) would be more suitable.

The recordings have given valuable information regarding forces acting on the head of the type of implant used. The design of the implant was intended to reduce bending loads on the stem and the operative technique (English, 1975) was intended to reduce the overall force acting across the hip joint by offsetting the trochanter and displacing the centre for rotation towards the mid-line.

One hundred and seventeen uninstrumented implants of similar design have been used, fifty-two with the

Table 2. Daily prosthesis force recordings

Load condition	Bodyweight	Days post operation 0	1	2	3	4	5	7	12	35	36	40
	2.5											
Supine	2.0											
	1.5											
	1.0											
	0.5											
	0											
Average force		0.3	0.7	0.7	0.8				1.0			
	2.5											
Sitting	2.0											
	1.5											
	1.0											
	0.5											
	0											
Average force						0.9	0.9	0.8	1.0			
	2.5											
Stand 2 legs	2.0											
	1.5											
	1.0											
	0.5											
	0											
Average force					*1.0	*1.0		*1.0	1.4	1.5	1.4	1.3
	2.5											
Stand 1 leg	2.0											
(L) instrumented	1.5											
(R) other leg	1.0											
	0.5											
	0											
Average force					*1.2	*1.3	*1.2		2.5		2.6	2.2
	2.5											
Walk	2.0											
(St) stance	1.5											
(Sw) swing	1.0											
	0.5											
	0											
Average force					*1.3	*1.4		*1.5	2.0	1.8	1.8	1.9
					*0.7	*0.7		*0.6	1.0	0.9	0.9	0.9

*Assisted activity (walking frame, stick)

Table 3. Comparison of results Rydell–English/Kilvington

Activity/Posture	Rydell Case 1	Rydell Case 2	Hull 12 days	Hull 40 days
Supine	Zero	Zero	1.02	
Sitting	Zero	0.2	0.98	1.0
Stand both legs	0.5	0.3	1.43	1.26
Stand gauged leg only	2.3	2.8	2.54	2.18
Stand other leg only	0.49	0.9	1.44	1.05
Walk				
Swing	0.85	1.2	1.01	0.9
Stance	1.8	3.0	2.0	1.8
Hip flexion 30°				1.2
Traction				
Flexed	0	0	0.93	
Extended	0	0	0.93	
Stairs up 15 cm				
Swing	0.1	0.3	1.02	
Stance	1.8	3.4	2.05	
Stairs down				
Swing	0.1	0.2	1.07	
Stance	1.8	2.6	2.06	

Force = × body weight

22 mm diameter head and sixty-five with the 32 mm diameter head.

The new shape stem is designed to be used in conjunction with the English socket which allows it to be precisely positioned in the pelvic wall. The risk of dislocation, fatigue failure and loosening is theoretically lower using this technique than with standard higher offset implants.

The very simple telemetry system and the unique design of prosthetic stem used to obtain these results are the first to be published since the work of Rydell in 1966. They are the first results of an *in vivo* measurement of human hip load using telemetric output. Comparison with Rydell's results shows considerable similarity (Table 3). Rydell found a stance phase force of 1.8 times body weight where we found 2.0 times body weight and he found swing phase forces of 0.85 times body weight where we found them to be 1.0 times body weight. Rydell found the one-legged stance caused a force of 2.3 times body weight in Case 1 and 2.8 times body weight in Case 2. These results are similar to our findings of 2.54 times body weight at twelve days after operation and 2.18 times body weight forty days after operation.

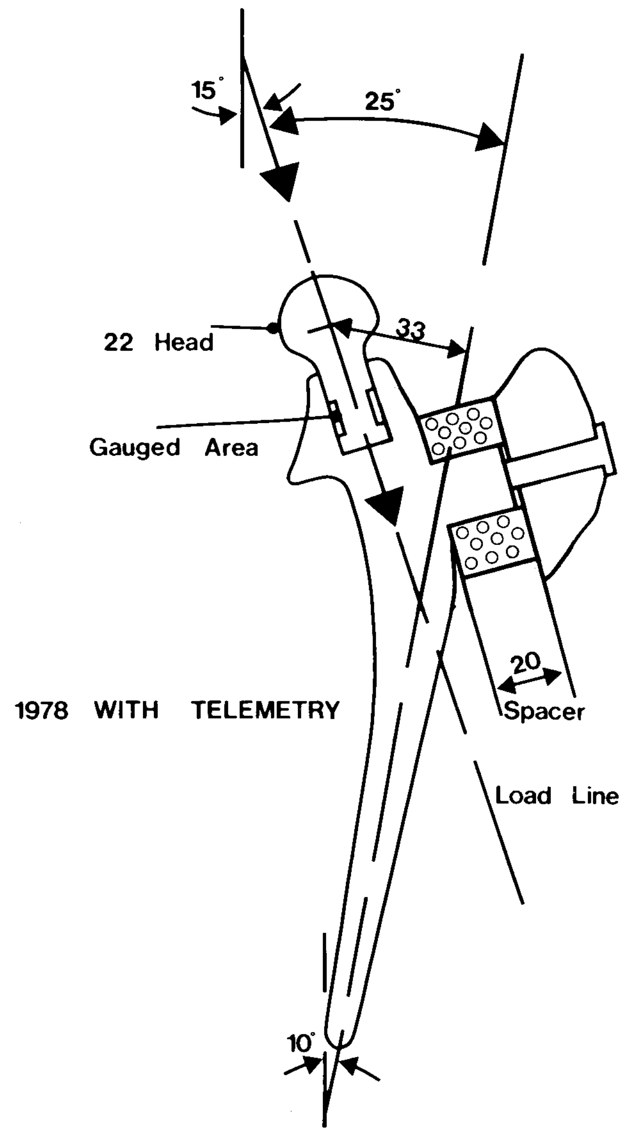

Fig. 16. English axial load prosthesis

Table 4. Range of leg movement thirty months post operation

	Right leg Standard English prosthesis (degrees)	Left leg Strain gauged prosthesis (degrees)
Flexion	100	95
Extension	-30	-30
IR	30	40
ER	20	10
Abduction	15	15

Differences were noted in comparison of forces during stair climbing and descending. Rydell recorded 1.8 times body weight in the stance phase and 0.1 times body weight in the swing phase when climbing (Case 1). In Case 2 he found 3.4 times body weight in the stance phase and 0.3 times body weight in the swing phase. His results were similar during descents. Our results show a somewhat higher force of 2.05 times body weight in the stance phase and 1.0 times body weight in the swing phase during climbing. Further comparisons with Rydell's results are difficult to make owing to the different principle used to measure the forces. Rydell's implant was gauged to measure bending forces in the implant neck set at 60° to the stem with an offset of 43 mm. Our system was designed to detect the mean axial force acting down the neck of the prosthesis, a very short distance away from the centre for rotation (Fig. 16). The implant was designed to have a neck aligned with the theoretical load axis in both planes computed by Paul and Morrison. Its offset was only 34 mm. Our method permitted records to be taken during implantation and throughout the post-operation recovery period, whereas Rydell delayed the connection of the buried wires attached to the implant gauges to his recording equipment for six months after operation.

Thirty-six months after implantation the patient remains problem free has no pain and undertakes all her usual domestic and outdoor pursuits (aged sixty). Her range of leg movement is detailed on Table 4.

It is not clear that symptoms may arise should corrosion occur in any of the sealed removable components. There is no clinical justification for removing these parts for examination at the present.

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