MECHANICAL SIMULATOR FOR THE UPPER FEMUR

E. MUNTING. M. VERHELPEN

This paper describes a simulator designed to evaluate the mechanical behaviour of the proximal femur and the influence of a prosthesis on this behaviour. Cyclic dynamic loads, corresponding to those generated by walking or several other activities, are applied to the proximal end of a fresh cadaveric femur before and after implantation of different types of hip prosthesis. Three musculoaponeurotic groups are modelled (abductor group, fascia lata and vastus lateralis). Load cells and strain gauges coupled to a data acquisition system are used to measure and record the forces on the femoral head and in the different muscle groups, as well as the strain variations occurring in the bone. Displacement transducers monitor the displacements occurring about the bone/implant interface.

Keywords : hip prosthesis ; biomechanics ; dynamic loading ; stability ; strain.
Mots-clés : prothèse ; hanche ; biomécanique ; mise en charge dynamique ; stabilité ; sollicitations.

INTRODUCTION

Total hip arthroplasty has become a routine procedure over the last 25 years. However there is still concern about implant loosening, especially in young patients. Whereas the results are very satisfactory and long-lasting in older patients with a lower level of a physical activity, in young patients loosening rates up to 50% within 5 to 10 years have been reported (7, 8, 11). Whereas wear and foreign body reaction against particulate material are probably the most important causes of long-term failure, mechanical factors are significantly involved in the loosening process in young patients. Therefore, besides research for the development of better articulating surfaces, implant design improvements for prevention of stress shielding, preservation of bone stock and vascularization and easier revision surgery is fully justified.

To evaluate the effect of a prosthesis on the strain pattern of the bone, many authors used the finite element method (FEM). By using an iterative procedure, Huiskes was even able to predict the bone response to stress shielding by FEM (13). Experimentally, the measurement of strains by strain gauges was frequently performed while the bone was under static loading. Oh chose different loads to simulate the forces acting on the femoral head during normal single-limb stance (16). Adams considered several instants on one cycle with a particular load reproducing the walking-loading cycle (1). Others used oscillatory joint loadings (3) sometimes following a sinusoidal waveform. Rarely has a specific attempt been made to replicate the complex dynamic loading patterns encountered in vivo. Jones developed an apparatus simulating the movements and the variable loads encountered in the hip joint (14). Because the mechanical behavior under dynamic loading may be totally different compared to static loading, a simulator designed to apply cyclic forces on the proximal femur was developed in order to evaluate the stress pattern generated in the proximal femur by any prosthetic replacement as well as the short-term stability of an implant.

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MECHANICAL SIMULATOR DEVELOPMENT

1. Proximal femoral loading during walking

Different mathematical models estimating the forces acting on the proximal femur during walking are presented in the literature. Pauwels assumed that slow walking can be considered as a succession of equilibrated one-leg supports (18). This is correct if the center of gravity moves in a straight line at a constant speed. In reality, the magnitude of the force on the femoral head in the case of one-leg support is smaller than in the case of walking, where the dynamic effect increases the loading because the centre of gravity moves in space. The acceleration of the centre of gravity in the vertical direction introduces dynamic forces increasing or decreasing the static loads. For a 1-Hertz walking rate it can be shown that the dynamic effect causes an increase of about 25% of the body weight (BW). In four studies, in vivo load measurments were performed. Rydell measured in vivo loads on two persons (19). During slow walking the resultant force on the femoral head is about 1.6 × BW and increases to 2.8 × BW during fast walking. The anteroposterior force represents 0.4 to 0.5 × BW. Otherwise a single-leg stance produces a force from 2.05 to 2.8 × BW. Brown measured on one patient a resultant head force of 2.19 × BW for single crutch walking and 2.05 × BW for an unsupported single-leg stance (4). Hodge, in one patient, obtained a force of 103% of BW (12) during the single-leg stance of normal walking. More recently, Davy obtained data on hip forces with a telemeterized total hip prosthesis implanted in one patient (9). During the stance phase of gait, the peak force was typically 2.6 to 2.8 × BW, with the resultant force located on the anterosuperior portion of the prosthetic femoral head. According to the in vivo studies the forces acting horizontally on the femoral head in the frontal plane have an anteroposterior direction. Some studies implying external measurments showed a force reversing between the two peaks of the cycle (17). The force/time curve in one cycle, according to Brown and Rydell's studies, has two peaks, the first being slightly greater (fig. 1). This curve was reproduced in our experiments with a maximum resultant force of 3 × BW, corresponding to fast walking. In accordance with Rydell and Brown, the same type of curve acts in the anteroposterior direction. As neither the flexor nor the extensor muscles are represented, it is appropriate to diminish the anteroposterior force from 0.5 to 0.2 times BW. In our model the pelvis equilibrium is ensured by the abductor, the vastus lateralis and iliotibial band, the resultant force on the femoral head and the partial body weight being produced by an actuator (body weight less one-leg weight). The anteroposterior force is equilibrated only by the elasticity of the femur.

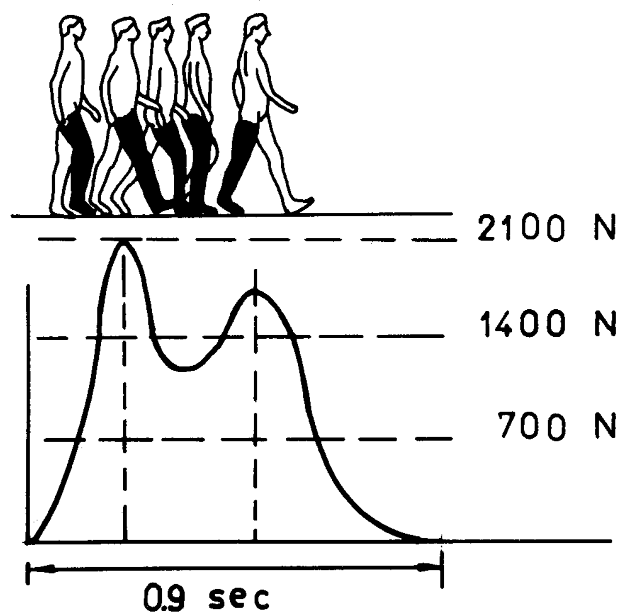

Fig. 1. — Force/time curve of the resultant force on the hip of a 70-kg man during normal walking.

2. Experimental set-up, muscle and pelvis modelling

Experiments are performed with fresh cadaveric femurs harvested within 24 hours of death. A radiogram of the bone is taken in order to exclude focal disease. In addition, a densitometric study is performed to accurately determine the mineral content. During the whole experiment the bone is kept wet by continuous irrigation with Ringer solution with 2% formalin. With the femur in physiological position, its condyles are embedded in polymeric cement contained in a cup secured to the base of the parallelepipedic frame of the simulator which is made of heavy steel girders secured to a thick steel plate, making it a very rigid construction. Three principal musculoaponeurotic strands are represented by cables : the abductor group (gluteus minimus and gluteus medius), the vastus lateralis and the iliotibial band. The proximal end of the cables representing the abductors and the iliotibial band are attached to one end of the lever arm representing the pelvis. The iliotibial band slips about the greater trochanter and is anchored on the base of the simulator like the vastus lateralis. The abductors and the vastus lateralis cables are further secured to a metal shell attached to the greater trochanter by means of polymeric cement. A polyethylene cup, whose position can be varied on the lever arm, constitutes the acetabulum and rests on the natural or prosthetic femoral head. The vertical force representing the body weight acts at the opposite end of the lever arm, whereas the anteroposterior force acts on the acetabular cup (figs. 2 and 3).

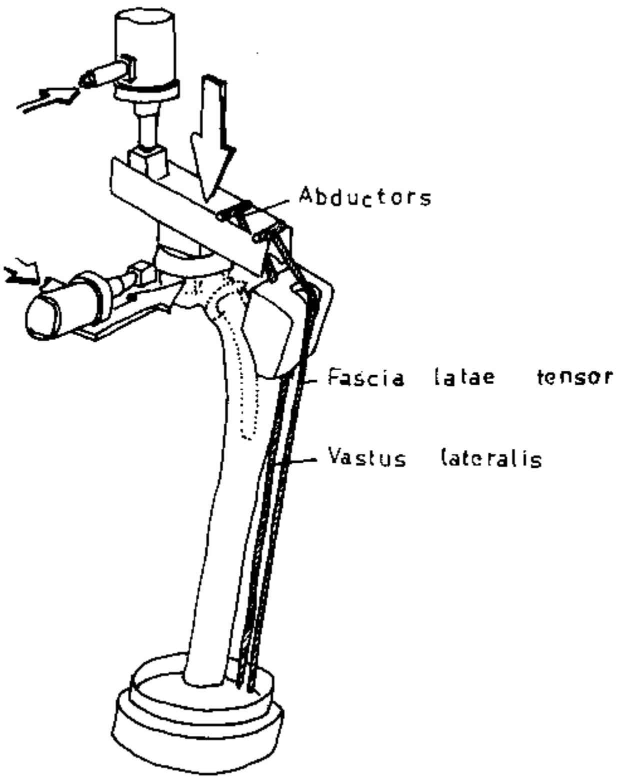

Fig. 2. — Schematic view of the simulator with muscles and the forces acting on the upper femur.

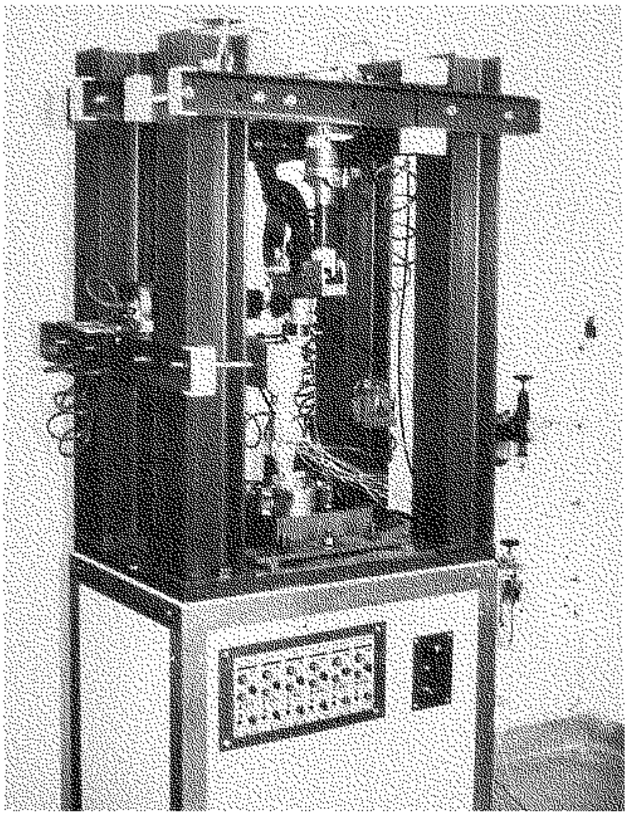

Fig. 3. — Framework of the mechanical simulator with a femur in place.

3. Dynamic loading apparatus

Two pneumatically powered actuators apply the vertical and anteroposterior forces. The loading curve during one cycle is obtained by alternately applying and stopping the pressurized air flow in the chamber behind the piston of the actuator, whereas a constant positive air pressure is maintained in the chamber located in front of the piston, allowing rapid withdrawal of the piston and thus unloading of the femoral head. Application and suppression of the pressurized air flow is obtained by an electromagnetic on-off valve controlled by an electric switch. The switch is activated by the signal of an emitting infra-red cell located in front of a rotating disc provided with black and white zones. A phototransistor is activated during the time that a white zone is passing in front of the infra-red cell. This turns on the switch and activates the solenoid of the air valve, setting the actuator in compression. Because of the compressibility of gases, the air pressure changes in the actuator need a response time which allows generation of a large variety of time/force curves by sucessively opening and closing the air valve (fig. 4). Not only can a slow or fast walking loading pattern be generated, but many other patterns can be created by modifying the black and white sequence on the disc. The frequency of the cycle can be varied by modifying the rotational speed of the disc and the maximum force by increasing the pressure of the air flow to one or both of the actuators.

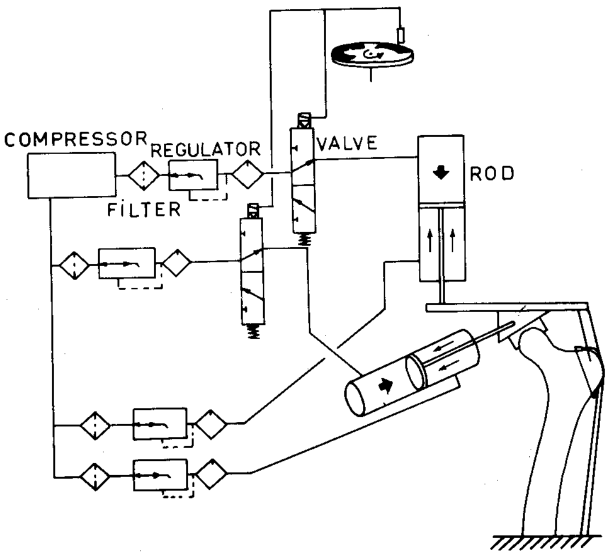

Fig. 4. — Schematic plan of the system simulating walking loading (→: continuous positive pressure; ➡: variable pressure).

4. Shock behavior

Shock behavior is evaluated by dropping a load of 28 kg from a precisely measured level, on a rod resting on the lever arm representing the pelvis. The magnitude of the impact is determined by the height of the fall. The load curve on the femoral head is studied by analyzing the signal of the load cell with an oscilloscope. The time to maximum loading is decreased to 30 msec, whereas the maximum load under walking conditions is reached in 200 msec. The influence of this parameter is interesting to stuy for a viscoelastic material like bone. This type of test is also useful to assess the fixation of an implant after a shock or in the failure mode, but accurate displacement measurement of the implant during the impact is not possible.

5. Force and strain measurements

Strain gauges are glued on the proximal and medial femoral cortex and at several levels around the upper part of the diaphysis : anteriorly, posteriorly, medially and laterally. This allows a comparative study of the strain distribution in the proximal femur before and after insertion of a prosthesis. The process of application of the strain gauges and their behavior in a wet saline environment has been studied previously (15) and in our own preliminary work (10). After scraping off the soft tissues, the bony surface where the gauge will be glued is defatted with ether and rubbed dry for 10 seconds with n° 320 emery paper. A neutralizer is then applied (M-Prep Neutralizer 5, Micro-Measurements, Romulus, Michigan). Methyl-2-cyanoacrylate adhesive (Cyanolit®) is used to glue the foil strain gauges (TML FLA-2, Tokyo Sokki Kenkyujo, Tokyo). Each gauge installation is sealed with a silicon layer (thermic dissipation paste) further covered with a wax layer (M-Coat W-1 Micro-Measurements). A 4-arm Wheatstone bridge circuit is constituted by two precision 120-Ohm resistors, one active gauge on the test bone and one temperature compensating gauge on an unloaded specimen. The signal is amplified (Strain amplifier 6M81, NEC San-Ei Instruments, Tokyo), digitized and recorded on a microcomputer. In order to start the measurement of each cycle precisely at the same moment, a second optical system detects a single signal on the disc and sends to the computer via the A/D converter an impulse initializing each cycle. About 300 values/sec are recorded to define the curve. Mean values are obtained from a series of 5 successive cycles with a standard deviation of less than 1% over a series of successive measurements. Electronic transducers measure the forces in the abductors, iliotibial band and vastus lateralis. Other transducers monitor the resultant forces on the femoral head in the frontal and anteroposterior plans. The force in the frontal plane has a 15° inclination about the vertical axis. In each situation (intact or implanted femur) identical head forces are easily obtained. Of course, if the centre of rotation of the prosthetic femoral head is displaced compared to the intact femur, the forces in the cables representing the muscles will be modified while the resultant force on the femoral head remains the same.

6. Measurement of bone-prosthesis stability

To measure the interface stability, a pair of displacement transducers (SENTEC, TYPE XL001.10F, LVDT) is secured to the prosthesis, successively oriented in the anteroposterior axis about the proximal and medial femoral cortex and in the frontal plane on each side of the upper aspect of the femoral neck. Each transducer reacts perpendicularly to its axis against a metal plate secured to the bone by sharp screws. For each pair of transducer's the mean value is calculated, corresponding to the displacement occurring between the bone and the implant at the level of the resection of the femoral neck. The transducers have a resolution of 0.001 mm through a range of 1.0 mm. These measurements, to be accurate and reproducible, need to be performed under static load conditions. In serial measurements, accuracy was found to be ± 2 microns as long as the transducer is kept in place. When removing and putting the transducer back in place between measurements, accuracy is significantly reduced : ± 10 microns. Mechanical comparators can be used instead with equal resolution and a constant accuracy of ± 1.5 micron.

7. Fatigue behavior of bone

Only elastic strains are measured by the strain gauges. From the effect of mechanical fatigue, dead bone, like many composite materials, exhibits a gradual loss of stiffness and strength well described by Carter *et al.* (6). This results in an increasing strain under an identical load. Bone being a viscoelastic material, creep occurs also during testing. Total deformation may be assessed by measuring the bone strains over a precisely set distance of at least several centimetres, with a mechanical comparator. For example, over a distance of 5 cm the elastic deformation under a load of three times body weight was found to be 190 to 200 microns in the beginning of a simulation, whereas this value increases to about 250 microns after 10^6 loading cycles. By leaving the comparator in place during the whole experiment, creep can also be measured.

DISCUSSION

A versatile tool has been developed to apply dynamic loading to the proximal femur. The simulator allows many different testing conditions (frequency and force variation, magnitude and proportion of tension in the different musculo-aponeurotic strands, static and dynamic analysis). The strain pattern in an intact femur may be compared to that generated by several different types of femoral components on the same bone. The effects of actions like walking, running, rising from a chair or climbing steps or even falling, on the stability of a prosthesis may be analyzed. Moreover, the evaluation of the stability of an implant under cyclic dynamic loading over 10^6 cycles, corresponding to several months of use in vivo, is of great interest since this stability is a determining factor in the formation of fibrous or bony tissue at the host-implant interface during the healing process (2, 5). The simulator allows determination of the relationship between the magnitude of the resultant force on the prosthetic femoral head and the interface mobility. Indeed, partial weight bearing during the healing process may prevent interface micromotion and allow bone ingrowth or apposition providing long-term stability under physiological loading.

Of course, the physiological reactions of bone to loading modifications or an unstable implant (reinforcing remodelling or resorption and osteolysis) do not occur in such an in vitro experiment. Furthermore the shock absorption and damping occurring under natural conditions by proprioceptive reactions are absent in this system, as well as the fine muscular balance uniting the pelvis with the lower limb. These facts make the simulation conditions even more severe. The short- and medium-term in vivo mechanical behavior of the bone-prosthesis construction may be reasonably extrapolated from the experimental observations obtained with this simulator.

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SAMENVATTING

E. MUNTING, M. VERHELPEN. Mechanische simulator voor het proximaal uiteinde van het femur.

Dit artikel beschrijft een simulator bestemd voor het evalueren van de invloed van een prothese op het mechanische gedrag van het proximale dijbeen eveneens als de stabiliteit van het implantaat op korte termijn. Verticale en horizontale krachten vergelijkbaar met die ontwikkeld gedurende het lopen of andere activiteiten worden gericht tegen het proximale uiteinde van een femur vóór en na het implanteren van verschillende heupprothesen. Drie spiergroepen zijn door kabels voorgesteld (gluteus medius en gluteus minimus, fascia latae, vastus lateralis). Krachtmetingapparaten en rekstrookjes zijn met een gegevensopneming systeem verbonden. Zo worden de krachten tegen de femorale kop en in de verschillende kabels die de spieren voorstellen gemeten. De vervorming in het bot wordt tegelijkertijd bestudeerd. Verplaatsingen tussen de prothesen en het bot worden ook gemeten.

RÉSUMÉ

E. MUNTING, M. VERHELPEN. Simulateur mécanique de l'extrémité supérieure du fémur.

Un simulateur mécanique a été conçu pour évaluer les déformations au sein d'un os intact ou en présence d'une prothèse. La stabilité mécanique à court terme de la prothèse est également étudiée. Une mise en charge dynamique composée, reproduisant les forces générées par la marche ou d'autres activités est appliquée à la partie supérieure d'un fémur frais avant et après implantation de divers types de prothèse. Trois groupes musculaires sont modélisés par des câbles (abducteurs, tenseur du fascia lata et vaste externe). Des capteurs de forces et des jauges de déformation, connectés à un système informatisé d'acquisition de données, permettent de mesurer les forces en présence et les déformations à la surface de l'os. Des capteurs de déplacement enregistrent les mouvements relatifs entre l'implant et l'os.