*Acta of Bioengineering and Biomechanics*
*Vol. 14, No. 2, 2012*

Original paper
DOI: 10.5277/abb120208

# Human pelvis loading rig for static and dynamic stress analysis

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This work is aimed at designing and constructing a loading rig for the synthetic hemi-pelvis; this system has been conceived with the goal of applying differently oriented articular forces in order to experimentally test the stress distribution and the stability of surgical reconstructions like, for example, hip arthroplasty or pelvic fixation. This device can be interfaced with a usual loading machine; it preserves the anatomy of the hemi-pelvis; it is simply constrained and it allows the simulation of all physiologic activities. Moreover, the visual accessibility of the peri-acetabular area has been guaranteed and this is imperative in order to be able to perform full-field analyses like a thermoelastic or photoelastic stress analysis.

First experimental trials have shown a good repeatability of loading–unloading cycles (<1.2%), a low hysteresis (<2.4%) and a good dynamic behaviour (up to 10 Hz loading frequencies).

*Key words*: *dynamical loading, hip arthroplasty, mechanical test, pelvic fractures, pelvis, thermoelastic stress analysis*

## 1. Introduction

Acetabular component loosening has been identified among the main limiting factors of the total hip arthroplasty (RITTER and GALLEY [18]); inadequate mechanical fixation or load transfer may contribute to the loosening process. In fact, the acetabulum cup changes the stress distribution with respect to the physiologic condition: a bone remodelling process is therefore initiated (MUELLER et al. [14]) which can eventually lead to acetabular loosening if some new equilibrium is not reached.

The experimental analysis of the stress distribution on the pelvis, produced by physiologic activities, is therefore a key factor for the design of an "optimal" cup.

One of the simplest loading rigs was introduced by FINLAY et al. [8]: the pelvis was constrained to the loading machine leaving all translational degrees of freedom; the femoral head and its neck were directed vertically and centred over the load cell of the testing machine. The strain distribution was mapped through twenty five strain gauges and photo-elastic stress analysis. MICHAELI et al. [13] adopted a loading system where various orientation of the articular force could be simulated and contact stresses were measured by means of a pressure-sensitive film. MASSIN et al. [12], WIDMER et al. [21], and RIES et al. [17] employed more complex loading systems where the whole pelvis (sacral vertebrae included in the first work) was analysed and muscle forces were simulated as well; the strain distribution was assessed by means of seven (MASSIN et al. [12]) to ten rosettes (RIES et al. [17]), while acetabular contact pressures were measured by means of pressure-sensitive films (WIDMER et al. [21]).

The authors intended to perform a full-field thermoelastic stress analysis (HARWOOD and CUMMINGS [9], DULIEU-BARTON and STANLEY [7]); this is a full-

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Received: October 13th, 2011
Accepted for publication: March 31st, 2012

field experimental technique which allows us to obtain the map of the sum of principal stresses, given a dynamic loading condition. A suitable loading rig needed to be designed, satisfying these main requirements:

• simple constraining in order to be able to apply repeatable, pre-determined loads,

• good results' repeatability and minimum hysteresis in order to achieve reliable, general results,

• versatility in order to simulate all possible physiologic activities (such as gait, raising and sitting on a chair, etc.),

• the possibility of applying sinusoidal loads (at least up to 5 Hz),

• a good visual accessibility.

The two last requirements are mandatory for full-field thermoelastic stress analysis. All those requirements induced us to sacrifice the simulation of muscular loads: their broad areas of attachment would seriously limit visual accessibility, and the need of cables and pulleys would compromise the dynamic response of the rig.

The engineered loading rig is illustrated in the following, having tested its performance and its suitability for a full-field experimental stress analysis. This rig has demonstrated its flexibility having been applied to the study of another kind of pelvic surgery that is the external fixation of a fractured pelvis.

## 2. Materials and methods

The loading rig has been designed to be interfaced with commercial testing machines (in this case Instron 8872); synthetic hemi-pelves (Sawbones Europe AB, Sweden) were used as test specimens.

The pelvis is constrained between two spherical joints in order to avoid an over-constrained structure and to be able to reproduce all the possible orientations of the articular force (figure 1). The lower spherical joint (LSJ) is made of a concave component which is screwed to the loading cell, while the convex component is connected to the pelvis through a prismatic block (P1, figure 1) and a resin block (B, figure 1) into which a part of the iliac wing is embedded. The upper spherical joint (USJ) is realised by the hip joint itself; it might be the physiologic joint or a prosthetic one, depending on the test of interest; the convex component (that is the femoral head or its prosthetic substitution) is screwed to the hydraulic actuator. The two joints (USJ and LSJ) share the same centre of rotation and this makes it possible to rotate the hemi-pelvis with respect to the loading vertical force; besides, the centre of rotation is the locus where all constraining forces pass.

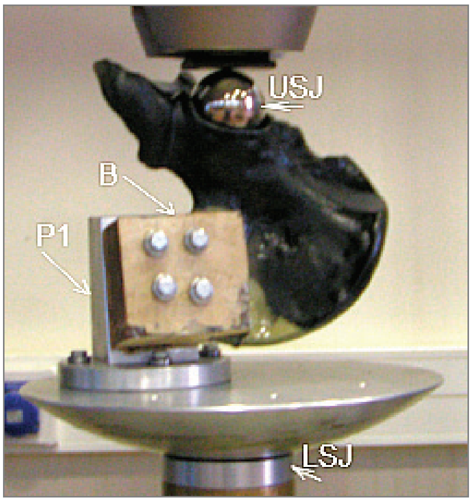


Fig. 1. Loading rig: the pelvis is constrained between two joints: the lower spherical joint (LSJ) and the upper spherical joint (USJ); the LSJ convex component is connected to the pelvis through a prismatic block (P1) and a resin block (B)

The physiological load directions have been obtained from the experimental studies by BERGMANN et al. [3] where a hip prosthesis was instrumented; they considered nine recurrent activities: walking slowly, walking normally, walking fast, stairs up, stairs down, chair up, chair down, standing on, and knee bend, and reported the value and orientation of the articular force for each activity. In figure 2, a 3D model of the hemi-pelvis was created from CT and the articular force vectors arising during drawing all the mentioned activities. The graphical representation demonstrates that simulating all activities would have required a very large angular excursion of the pelvis with reference to the LSJ (over 90 degrees). A different solution was here sought since the authors observed that all vectors could be enclosed by two distinct cones whose angular aperture was less than 50° (figure 2). Consequently, two different fixtures ($P_1$, $P_2$ in figure 3) for the pelvis have been designed: each fixture allows the alignment of the articular force with the axis of one of the aforementioned cones, when the lower spherical joint is in its centred position. Therefore, the lower spherical joint could be designed to allow only ±25° angular excursion (figure 4).

Two inclinometers were placed in the frontal and in the sagittal planes, on the moving component of the LSJ to measure the spatial inclination of the hemi-pelvis.

The simulation of a given physiologic activity can be so accomplished, following the following steps:

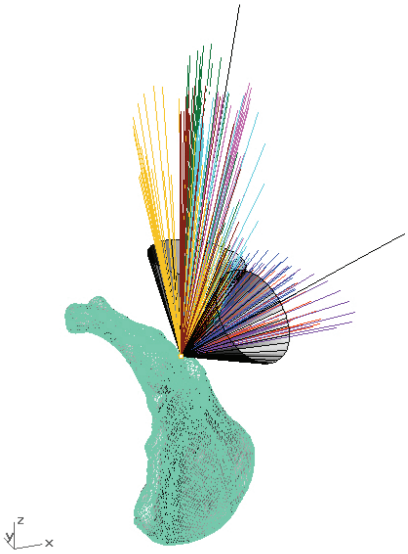

Fig. 2. Articular forces produced by daily activities;
all vectors can be enclosed in two cones

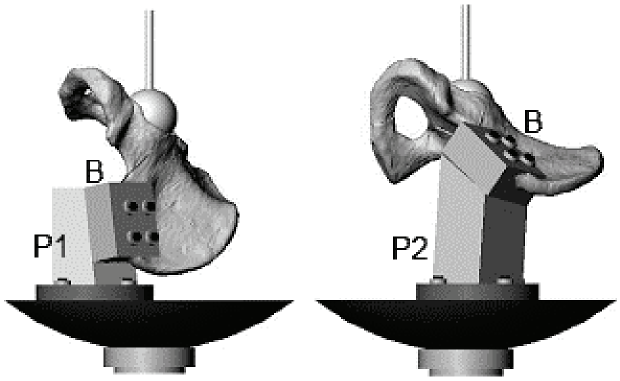


Fig. 3. Two assembly modes;
the block connected to the pelvis (B) remains the same,
while the prismatic block screwed to
the spherical cup changes ($P_1$ or $P_2$)

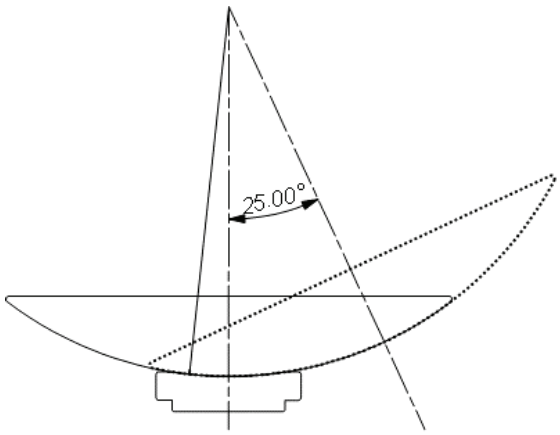


Fig. 4. Range of motion of the lower joint (LSJ)

• to specify the cone in which the selected activity is enclosed;

• to select the appropriate prism;

• to assembly the prism to the LSJ and to the hemi-pelvis;

• to rotate the hemi-pelvis in order to reach the correct position.

A strain gauge rosette was bonded to the hemi-pelvis in the periacetabular area, just below the contact area between the spherical head and the cotyle; strain gauge measurements and load–displacement signals given by the testing machine have been used to assess the loading rig performance.

The synthetic pelvis was black painted in order to perform a thermoelastic stress analysis. Full-field thermoelastic maps were acquired using a standard infrared thermocamera (ThermaCam SC3000 by Flir Systems,Wilsonville, OR: 320×240 pixesl; 50 Hz frame rate; 20 mK thermal resolution), and an appositely developed software (ZANETTI et al. [22]). According to thermoelastic theory (HARWOOD and CUMMINGS [9], DULIEU-BARTON and STANLEY [7]), temperature maps can be converted into the maps of the first stress invariant:

$$\Delta T = -K_m T \cdot \Delta I_{1,\sigma},$$

where:

$K_m$ is the thermoelastic constant of the material (measured through calibration),

$T$ is the absolute ambient temperature,

$\Delta I_{1,\sigma}$ is the first stress invariant,

$\Delta T$ is the amplitude of temperature variation, measured by the differential thermocamera.

The applicability of this technique to the constitutive composite material of synthetic bones has been demonstrated in our previous work (ZANETTI and AUDENINO [23]), where the calibration constant has been calculated, considering uniaxially stressed specimens ($K_m$ = 0.038 GPa$^{-1}$).

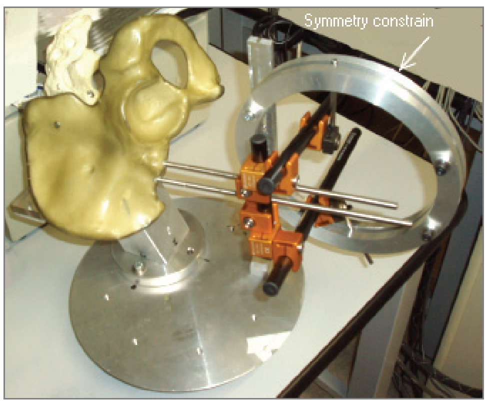


Fig. 5. Testing set up for a pelvic external fixator

A new application has been implemented in order to demonstrate the flexibility of the rig designed (figure 5): the stability of an external fixation device has been studied, acquiring load/displacement signals: in this case, the hemi-pelvis is in contact with the sacrum bone (simulating a fracture); the iliac bone is fully constrained to LSJ; half an external fixator construct is applied, and a symmetry constrain has been added.

# 3. Results

The performance of the loading rig has been assessed through repeated loading cycles (ANSI/ISA [1]), with a compressive vertical force between –400 N and –2000 N at 10 Hz. Ten cycles were given at the beginning for settlement, and, after, ten cycles have been completed at 200 Hz sampling rate.

Sample curves depicting the highest principal strain are plotted in figure 6; the repeatability is good: given a certain load and a certain direction (loading/unloading cycles), the maximum strain error is equal to 0.003 με that is 1.2% of the maximum measured strain.

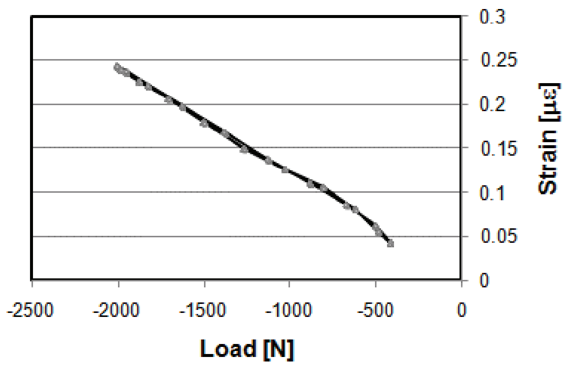


Fig. 6. Strain–load curves for a natural joint and a prosthetic joint

The linearity error (ANSI/ISA [1]) has been evaluated considering the average cycle and measuring the largest deviation of strain from the terminal line (that is the line connecting the endpoints corresponding to –400 N and –2000 N loads, respectively); linearity error is equal to 0.0024 με that is 1.0% of the maximum measured strain.

The slope of the strain $\varepsilon$ vs. load curve is –1.21E-4 με/N with a maximum variability of 0.7%; it should be noticed that this slope is strongly dependent on the congruence of head–acetabulum coupling (MARKOLF and AMSTUTZ [11]).

The hysteresis loop area is acceptable: in the worst case, i.e., where the physiological joint is simulated, the hysteresis error (ANSI/ISA [1]) does not exceed 0.006 με that is 2.4% of the maximum measured strain; this is a good result, considering the irregular shape of both components.

Preliminary thermographic maps have been acquired (figure 7); a stress peak (3.8 MPa) is evident just below the contact area; when tests were repeated, the standard deviation of this stress peak reached 2.8%.

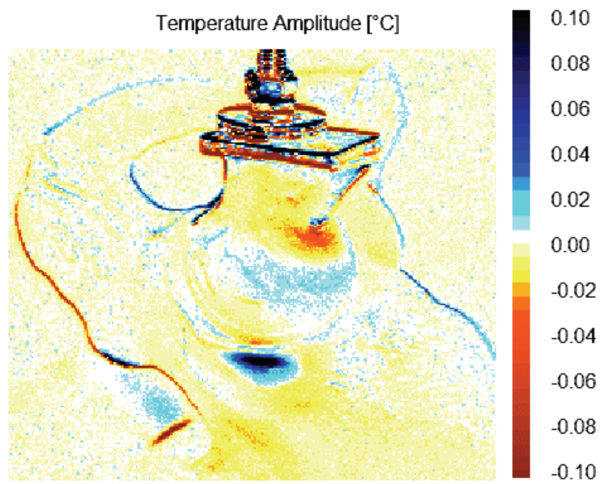


Fig. 7. Thermoelastic map

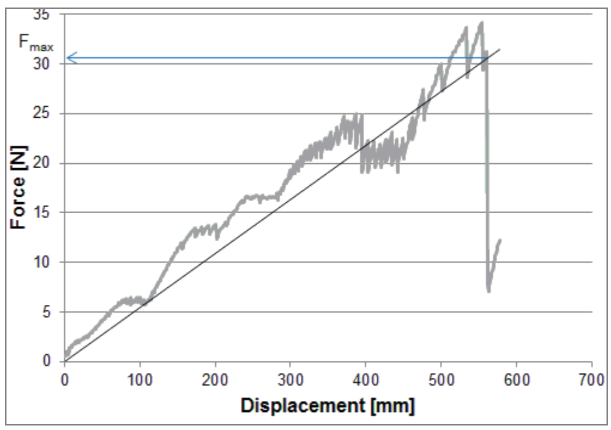


Fig. 8. Force-displacement curve, obtained on a pelvic fixator; $F_{max}$ is the dislocating force

The simplicity of this system has allowed its implementation to the study of another surgery that is the external fixation of a fractured pelvis; figure 8 presents a typical force/displacement curve; the experimental set up in figure 5 allowed us to establish a maximum dislocating load equal to 32 N, with an accuracy of 12%.

# 4. Discussion

In literature, various loading systems have been introduced for the pelvis (FINLAY et al. [8], MASSIN et al.

[12], WIDMER et al. [21], RIES et al. [17]); most of these were designed in order to simulate only one specific loading condition; moreover some of them require a whole pelvis as a specimen, sacral vertebrae included (SPARKS et al. [19]). The loading rig developed by MICHAELI et al. [13] is similar to the one introduced here for what concerns the underlying principle; however, the overall size and mass have been here greatly reduced thanks to the detailed observation of the actual force orientation and to the design of two alternative fixtures; this aspect proved to be relevant for what concerns the dynamic response of the loading set up.

Some authors (WIDMER et al. [21], RIES et al. [17]) focused on the accurate simulation of the main muscular forces; in this work, this aspect was sacrificed in favour of device simplicity, loading repeatability, dynamic response, and good visual accessibility. However, the effect of muscle forces is obviously taken into account through its resultant that is the articular contact force; DALSTRA and HUISKES [6] demonstrate how this approach does not hamper the obtainment of sound results in the peri-acetabular area, the latter being far enough from muscle attachment points. Besides, an accurate and reliable loading condition, even if simplified, can better detect small changes in stress distribution induced by a prosthesis, compared to more complex, even if more realistic, loading conditions.

Experimental tests have been performed on synthetic femora due to difficulties in finding, handling, and preserving real bones. Moreover, the physical properties of synthetic bones are similar to those of real bones and provide lower variability in testing (CRISTOFOLINI et al. [5]).

The simplicity of this experimental set up has made it suitable to the application of dynamic loads, as required by thermoelastic stress analysis. The power of thermoelastic stress analysis method lies in the possibility of obtaining full-field experimental data: the alternative application of strain gauges would be very onerous: more than twenty strain-gauge rosettes were used by RITTER and GALLEY [18] in a similar experiment. In biomechanics, thermoelastic stress analysis has been seldom used, mostly in a qualitative manner; VANDERBY and KOHLES [20] are the authors of one of the first biomechanical applications in uniaxially loaded cortical bone cubes; more recently, the same authors, i.e., KOHLES and VANDERBY [10], applied thermography to canine femora; ZANETTI and AUDENINO [23] carefully explained under which conditions differential thermography could be applied in generic, anisotropic materials; REFIOR et al. [16] employed fresh specimens and tested implanted femora; in this case, the surfaces could not be coated with black paint, a major shortcoming in the experimental procedure, since emissivity is low and cannot be regarded as constant over the entire specimen.

The stress distribution obtained by us through thermoelastic stress analysis is similar to other reported in literature, obtained by means of numerical methods (CILINGIR et al. [4]); a more detailed comparison would require the knowledge of the full first stress tensor invariant rather than von Mises stress, since thermoelastic stress analysis gives the sum of principal stresses.

As demonstrated, this loading system is versatile and well suited to the experimental simulation of various physiologic activities; for example, a pelvis fixation construct has been tested. The data obtained are consistent with those obtained by other authors who tested a "type C" fracture (PONSON et al. [15], ARCHDEACON et al. [2]): these authors obtained an ultimate unipodal load ranging from 10 to 43 N (PONSON et al. [15]) or from 49 to 64 N (ARCHDEACON et al. [2]).

## Acknowledgements

Many thanks to Dr. Massimiliano Salaorno for his advice and collaboration in the experimental set up.

This work was funded by the Italian Ministry of Instruction, University and Research (MIUR).

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