Biomechanics includes research and analysis of the mechanics of living organisms and the application of engineering principles. In this seminar, a determination of the contact stress in the human hip joint is presented. Two different methods are outlined. The first method involves in vivo measurements by a specially designed hip endoprosthesis. While the second method is based on a simple mathematical model, which uses as input data, a standard anteroposterior radiograph of both hips and pelvis. Conclusions obtained by using both methods indicate, that hip biomechanics is useful in designing the optimal treatment of the diseased hip as well as the rehabilitation process.
Contents

1 Introduction
2 Human hip joint
3 Hip stress measurement
   3.1 The instrumented prosthesis
   3.2 Contact pressures in the hip joint measured in vivo
4 Model of hip forces and stress distribution
   4.1 Resultant hip force
   4.2 Stress distribution
5 Conclusion
References
1 Introduction

Degenerative processes in the hip leading to coxarthrosis (a degradation of the hip cartilage) affect a population of elderly in the developed world. This population is increasing in number since its life expectation increases. Coxarthrosis can occur secondary to hip disorders such as necrosis of the femoral head or dysplasia of the hip. In hips with necrosis of the femoral head, a part of the head decays and in an advanced stage of the disease cannot bear weight. In hip dysplasia, malformations of the femur and the acetabulum (such as not having enough lateral coverage of the femoral head by the acetabulum) are present. In both cases, the weight bearing area is decreased leading to an increase of the contact hip stress and early coxarthrosis. Such conditions are treated by osteotomies intended to redistribute and diminish weight and also by implantation of the hip endoprosthesis.

Measurements of the hip stress are important to optimize the treatment and recovery of patients. Here we present two methods for determination of the contact hip stress. The direct measurement with a specially designed endoprosthesis yielded information on the general effect of particular activities on hip stress [1, 2, 3, 4]. This method can not be used in clinical practice where the method should be noninvasive, simple, and not too expensive. Therefore, simple methods based on mathematical models are more appropriate. We briefly present the method HIPSTRESS for determination of the contact hip stress distribution [5, 6, 7].
2 Human hip joint

The hip joint is a synovial joint formed by the articulation of the rounded head of the femur and a cup-like acetabulum of the pelvis. It is classified as a ball and socket joint. It forms the primary connection between the bones of the lower limb and the axial skeleton of the trunk and pelvis. Both joint surfaces are covered with a strong, but lubricated layer, called articular hyaline cartilage. Figure 1.

3 Hip stress measurement

Forces and stress measurements \textit{in vivo} were performed by a special prosthesis with implanted measuring devices\cite{4, 1, 2, 3}. The devices were mounted into the prosthesis stem and into the joint contact area of the prosthesis\cite{6, 8}. Such prosthesis gives us direct data, but such measurements can only be made on patients which need a prosthesis and therefore the data is only partially equivalent to healthy hips.

A miniaturized radio telemetry device has been developed which can monitor the magnitude and distribution of stress generated between cartilage-covered surface of human hip socket and the surface of a hip prostheses which replaces the spherical head of the femur in the hip\cite{4, 1, 2, 3}. The prosthesis consists of several pressure transducers and the telemetry system using an external power source.
3.1 The instrumented prosthesis

The instrumented prosthesis contains two principal components – a network of 14 identical pressure transducers and a miniaturized telemetry device.

Each pressure transducer consists of two main elements – a diaphragm which converts the applied pressure into a mechanical motion, and a displacement sensing device which measures the diaphragm motion and produces an electrical signal proportional to the diaphragm motion. A cross section of a complete transducer is shown in Figure 2. The diaphragm is circular and is machined directly into the wall of the prosthesis ball. In this manner the sphericity of the outer surface of the ball is preserved and there are no edges or holes that might damage the cartilage surface in the acetabulum. A sliding pin held in plastic bushing couples the motion of the centre of the diaphragm to the free end of a cantilever beam. The cantilever beam is a silicon crystal which has four semiconductor strain gauges diffused into its surface, which sense the bending of the beam and produce an electrical signal in response to the pressure applied to the diaphragm.

The pressure transducer is designed to operate between 0 and 690 N/cm² and has sensitivity of approximately 10 \( \mu V cm^2 V N \). The natural frequency of the transducer is approximately 10 kHz, and the strain gage bridge is a linear function of pressure to within one percent of the full-seal output. Temperature stability of the transducers is not a problem, partly due to the inherent temperature compensation of the four-active-arm strain gage bridge, but primarily because the transducers will operate in temperature controlled environment (the human body).

A full complement of 14 transducers mounted into prosthesis hemisphere is shown in Figure 3. The diameter of the hemisphere is 4.8 cm.

A radio telemetry device is incorporated into the instrumented prosthesis to eliminate the possibility of infection that would exist if wires were brought out through the experimental subject’s skin. With the exception of a small coil mounted on the end of the prosthesis stem, the implanted telemetry...
device is completely enclosed and hermetically sealed within the ball of the prosthesis.

The telemetry device contains no batteries – it is totally powered by a source external to the subject’s body by means of a transformer power-induction link. This approach to powering the telemetry device helps solve two design problems – device size and operating lifetime. The availability of relatively large amounts of power at relatively high voltages (several tens of volts) allows considerable freedom in terms of component selection and circuit design. In addition, the power induction system eliminates several problems associated with batteries, such as low energy storage density, externally operated switches, and the potential toxic hazards of batteries used in long-term implantations.

3.2 Contact pressures in the hip joint measured in vivo

A pressure-measuring prosthesis was implanted in a 68-year-old patient who had sustained a displaced fracture of the femoral neck [2]. The data reveal very high local (up to 18 MPa) and nonuniform pressures, with abrupt spatial and temporal gradients [2]. Figure 4 shows locations of transducers.
Figure 4: Drawing of the endoprosthesis with pressure-measuring instrumentation showing the locations of the transducers. From [2].

<table>
<thead>
<tr>
<th>Activity</th>
<th>Pressure (MPa)</th>
<th>Transducer number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Postoperative activities</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Balanced suspension</td>
<td>0.7</td>
<td>7</td>
</tr>
<tr>
<td>Supine (bed rest)</td>
<td>1.4</td>
<td>5</td>
</tr>
<tr>
<td>Buck traction (22 N)</td>
<td>1.5</td>
<td>5</td>
</tr>
<tr>
<td>Riding stationary bicycle</td>
<td>1.6</td>
<td>5</td>
</tr>
<tr>
<td>Use of continuous passive motion</td>
<td>1.7</td>
<td>5</td>
</tr>
<tr>
<td>Abductor splint</td>
<td>2.6</td>
<td>10</td>
</tr>
<tr>
<td>Use of bedpan</td>
<td>3.2</td>
<td>5</td>
</tr>
<tr>
<td>Resisted isometric...</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip adductors</td>
<td>5.0</td>
<td>5</td>
</tr>
<tr>
<td>Hip extensors</td>
<td>4.9</td>
<td>5</td>
</tr>
<tr>
<td>Hip abductors</td>
<td>4.2</td>
<td>5</td>
</tr>
<tr>
<td>Hip flexors</td>
<td>3.5</td>
<td>5</td>
</tr>
<tr>
<td>Everyday activities</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rising from a 38 cm high chair</td>
<td>15.0</td>
<td>1</td>
</tr>
<tr>
<td>Rising from a 56 cm high chair</td>
<td>9.2</td>
<td>1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Activity</th>
<th>Pressure (MPa)</th>
<th>Transducer number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mobilization</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Parallel bars</td>
<td>3.4</td>
<td>5</td>
</tr>
<tr>
<td>Walker</td>
<td>3.8</td>
<td>5</td>
</tr>
<tr>
<td>Partial weight</td>
<td>3.5</td>
<td>5</td>
</tr>
<tr>
<td>Bearing crutches</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cane – 65 N</td>
<td>5.1</td>
<td>3</td>
</tr>
<tr>
<td>Cane – 130 N, 220 N</td>
<td>4.8</td>
<td>3</td>
</tr>
<tr>
<td>Unsupported</td>
<td>5.5</td>
<td>3</td>
</tr>
</tbody>
</table>

Table 1: Peak stress in different activities and transducers. From [2]
This table 1 shows the maximum pressure registered on the transducer that had the highest reading during the activity at that particular testing session. The largest maximum pressures occurred during isometric adduction and extension of the hip [2]. Isometric muscle contractions caused maximum pressures in the hip that were nearly equal to those generated during normal walking.

To initiate walking postoperatively, the patient started by using parallel bars and progressed to using a walker, to non-weight-bearing with crutches, to partial weight-bearing with crutches, to using a cane, and finally to unsupported gait. Progress through this process of rehabilitation did not always result in progressively increasing pressures. Rather, non-weight-bearing with crutches produced the lowest maximum pressure – a reduction of 56 per cent compared with the maximum pressure during normal walking. Partial weight-bearing on crutches reduced the pressure by only 35 per cent. A single cane in the opposite hand, when loaded at 130 N, produced a reduction in maximum pressure of only 11 per cent compared with unassisted walking. When an additional 90 N of force was applied to the cane, maximum pressure was not further reduced. The use of ischial-bearing brace (a brace strapped onto upper part of the leg and knee, that bears the lower part of the pelvis) was as effective as partial weight-bearing with crutches in reducing maximum pressure.

During the gait cycle, the pressure was typically maximum at transducer 3, which articulated with the superior acetabular dome. This shows that the pressure distribution has its pole in the superior region and its size corresponds with the weight applied to the leg.

As rehabilitation proceeded, measurements were recorded for the more strenuous activities of rising from a chair, stair-climbing, jogging and jumping. Rising from a chair produced a maximum pressure that was approximately three times that which occurred during walking and approximately twice that occurring during jogging or jumping. The maximum pressure decreased as the height of the chair increased. The greatest maximum pressure was 18 MPa on transducer 1, when the patient rose from a normal chair (45 cm high) twelve months postoperatively.

These measurements suggested that postoperatively, using crutches without bearing weight and using a stationary bike is beneficial for the patient since the motion at low hip stress is enabled. On the other hand, rising from a low position was not advised. The results of the direct measurements indicated practical advice which helped improve the treatment.
4 Model of hip forces and stress distribution

The advantages of mathematical modelling over direct measurement is non-invasiveness and making it possible to study large groups of hips. Input data for such models are, for example, geometrical parameters of the body, accelerations of certain points of the body and external forces. The shortcomings of the mathematical models are that simplifications are introduced, therefore their accuracy is limited.

The hip load can be described with a vector sum of all forces, that are transferred from the acetabulum to the head of the femur (this sum defines resultant hip force $\vec{R}$) and with the distribution of this force over the load bearing area – e.g. pressure that acts on the hip. $\vec{R}$ can be either measured directly or calculated via a mathematical model combined with external measurements.

4.1 Resultant hip force

In the biomechanical description, the body is divided into segments connected with joints. Each segment is described as a rigid body of a certain mass ($m$), center of gravity ($G$) and rotational momentum ($J$). The movement of the segment is caused by forces which act on it: its weight $\vec{W}$ and forces of other segments (inter-segment forces). The inter-segment force is a vector sum of all forces, which act on the joint surface and forces which are caused by anatomical structures (t.i. muscles, ligaments). The contact point of the inter-segment force is at the connection between two segments. Such forces for a any chosen segment are shown on Figure 5.
Figure 5: Forces that act upon a chosen segment. $C$ is the rotational axis, $G$ is the centre of mass, $\vec{W}$ weight, $\vec{F}_{11}$ and $\vec{F}_{12}$ are the inter-segment forces of the first and the second segment acting upon the chosen segment. From [9].

The segment dynamics are described by equations of motion:

\[ \vec{W} + \sum_i \vec{F}_{1i} = m\ddot{\vec{a}}, \]  
(1)

\[ (r_{CG} \times \vec{W}) + (r_{CG} \times m\ddot{\vec{a}}) + \sum_i (r_{CIi} \times \vec{F}_{1i}) + \sum_i M_{1i} = J\ddot{\vec{\alpha}}, \]  
(2)

where $\vec{\alpha}$ is angular acceleration of the segment, $r_{CG}$ is a vector from the rotational axis ($C$) to the centre of mass ($G$) and $r_{CIi}$ is a vector from rotational axis ($C$) to the point where inter-segmental force of segment $i$ acts on the chosen segment, $M_{1i}$ is the inter-segmental torque, of the segment $i$ on the chosen segment. Index $i$ runs over all segments which act on the chosen segment.

In particular, when describing forces acting on the hip, we mark one leg as a first segment and the rest of the body as the second segment.

The inter-segment force which acts between two segments is equal to the sum of the joint force $\vec{R}$ and the force exerted by the muscles $\vec{F}$ [6]. Here, ligament forces are neglected.

\[ \vec{F}_i = \vec{R} + \vec{F}. \]  
(3)

To estimate the resultant hip force $\vec{R}$, we have to know the inter-segmental and muscle forces. If we were to include the forces which act upon the
segment into the model, the number of unknowns would exceed the number
of equations. Several methods for solving this problem were introduced, one
of them based on the reduction of the number of unknowns (e.g. neglecting
the forces of ligaments and tendons).

The resultant hip force $\vec{R}$ can be estimated by using weight and geo-
metry data obtained from anteroposterior radiographs. By determining some
characteristic points on the pelvis and the femur, the muscle insertion points
are estimated by re-scaling the reference points determined on a cadaver [7].

4.2 Stress distribution

Since the resultant hip force acts over the load bearing area of the hip, we
would also like to describe its spatial distribution – i.e. the stress distribution
over the load bearing area.

$$\vec{R} = \int p \, d\vec{A},$$

where $p$ is stress and $d\vec{A}$ is the area element. Within a simple model Hip-
stress it is assumed that $p = p_0 \cos \gamma$, where $p_0$ is stress at the stress pole, $\gamma$
is the angle between the radius vector from the centre of the femoral head to
the chosen point on the femoral head and the radius vector from the centre
of the femoral head and the stress pole. The cosine function implies perfect
spherical shape of the femoral head and acetabulum and the validity of the
Hooke’s law for the deformation of the soft cartilage between bony acetab-
um and the femoral head upon the loading of the hip. The integration is
performed over the load bearing area, which is defined on the lateral side by
the coverage of the femoral head by the acetabulum and on the medial side
by the condition of vanishing stress.

The Hipstress model was used to analyze different populations of hips.
When estimating a resultant hip force, a representative position (the one
legged stance) was used.

The results showed that an elevated hip stress represents a risk factor for
development of coxarthrosis: in patients with higher peak stress the implanta-
tion of the endoprosthesis occurred at a younger age (Figure 6). Elevated hip
stress produces a less favorable environment, which speed up degradation of
the cartilage.

The model was also applied to the partial and total hip endoprosthesis.
In total hip endoprosthesis, the friction between the ball and the socket of
the prosthesis produces debris which causes microinfection at the prothesis-
tissue interface leading to the deterioration of the bone and eventually the
wear of the prothesis. By using the HIPSTRESS method, a positive correlation
between the maximal contact hip stress and a linear wear of the polyethylene cup, was found ten years postoperatively (Figure 7).

Figure 6: Dependence between age at implantation of hip endoprosthesis and hip stress
(B.Pompe, I.Rigler, I.Ratoša, R. Vengust, V. Antolič, A. Iglič, V. Kralj-Iglič, not published)

Figure 7: Correlation between maximal stress and linear wearing down of polyethylene cup used in prosthesis
(R. Košak, M. Daniel, V. Antolič, A. Iglič, V. Kralj-Iglič, not published)

The HIPSTRESS method proved to be useful in clinical practice for determination of the risk factors for coxarthrosis, for determination of the hip dysplasia and for determining the optimal position of the endoprosthesis as to minimize the production of wear debris.
5 Conclusion

Hip biomechanics is useful in planning the treatment and rehabilitation of diseased hips. Using instrumented prosthesis hip stress was measured in vivo during rehabilitation and later on during everyday activities. Experimental data showed the need to change the rehabilitation routine for patients with such prosthesis. Unfortunately, due to ethical standards, such measurements can rarely be done in vivo, and therefore statistical representation is extremely small. On the other hand simple methods based on mathematical models are non-invasive and can help analyze larger populations. Such analysis verifies their correctness and enables those methods to be used to determine risk factors for diseased hips and to plan the treatment.
References


