FORCES ACTING ON THE FEMORAL HEAD-PROSTHESIS

A Study on Strain Gauge Supplied Prostheses in Living Persons

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by

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"I often say that when you can measure what you are speaking about and express it in numbers you know something about it; but when you cannot measure it, when you cannot express it in numbers, your knowledge is of a meagre and unsatisfactory kind."

(*Lord Kelvin:* Popular lectures and addresses. Vol. 1 p. 80 Macmillan 1891)

Introduction

Numerous attempts have been made to determine the magnitude and direction of the forces acting on the hip-joint. The theories that shape and internal structure of the femoral head and neck are related to stress and strain, the mechanism of femoral neck fractures and the frequently occurring failure of its internal fixation have led to analyses of the forces involved. It is believed that pathological conditions of the hip are influenced by mechanical factors and many reconstructive procedures in this region are based on mechanical considerations.

At present all our knowledge of the forces acting on the hip-joint comes from theoretical calculations and tests on specimens and models. Most of the results are based on 2-dimensional studies.

The use of theoretical calculations has its limitations and direct measurements by means of strain gauges in the hip under living conditions is another way to get information about forces acting on the hip-joint. Recordings from strain gauges applied to living bone are at present not reliable due to the physical properties of bone. Therefore metal prostheses, substituting the upper femur, with built in strain gauges were designed and manufactured. These prostheses permit measurements of forces present to be determined as to its magnitude and direction.

The results obtained with these prosthesis are only valid for the prosthetic hip, but they could provide information applicable to normal hips if factors such as lever arms and directions of muscular pull are known about the normal and prosthetic hip. Recordings can be made from a patient with such a measuring prosthesis under different activities involving the hip such as standing, walking, running etc.

For the analyses while walking, a device describing the forces between the foot and the ground in consecutive steps was desirable. Therefore two electronic walk-ways were constructed, one for the left foot and one for the right.

This work has been divided in 5 parts. In the first part, historical reviews of studies on the shape and internal structure of the proximal femur, force analyses and the development of force-plates are presented. The second and third parts discuss the construction of the measuring prosthesis and the electronic walk-ways. The intravital measurements and the results are reported in the fourth part and in the fifth part the investigation is recapitulated discussed.

I. Survey of the literature

Structure of the proximal femur

In his work on strength of material Galilei (1638) discussed the loadbearing capacity of bones. Later, many others dealt with the same subject, among them, Ward (1838), Wyman (1850), Haughton (1864), Meyer (1867), Wolf (1870—1900), Roux (1893—1896) and Pauwels (1948—1958). They are of the opinion that there is a connection between design and function of the bone.

The design of the proximal femur, the head and the neck, and its internal architecture has been the subject for mechanical analyses.

The hip-joint is a ball and socket joint. The head of the femur is not a perfect sphere. It is slightly compressed in an approximately ventro-dorsal direction. The difference between the two principal axes is very small, the ratio between them being 1.02.

The femoral neck lies in a distal-lateral direction from the head. It is a 30—40 mm long tube and, like the head, is compressed in an approximate ventro-dorsal direction. The neck changes its shape in its course. At its junction with the head, a section through the neck is almost cylindrical. The shape of neck increasingly becomes elliptical in a distal-lateral direction. In the middle of the neck, the ratio between the two principal axes is 1.15 and at the junction between the neck and the femoral shaft the ratio is 1.65 (Backman 1957).

The direction of the major axis of the elliptical section does not have the same direction as the long axis of the femoral shaft. The major axis forms an angle ω_1 opening dorsally and pointing downwards (fig. 2) in relation to the shaft. The cortical shell of the neck is as thin as paper at its head end. As the shaft is approached the cortical bone increases its thickness gradually. In the superior part of the neck the increase is small, but in the inferior part, the thickness increases considerably and is maximum where the major axis intersects the cortical bone. Therefore in a section through the major axis the cortical bone is strongest in the inferior part. It is of interest to note that a newborn femur has cylindrical neck and during growth the elliptical form develops.

To be able to understand the design of the upper femur and its relationship to the forces acting on the bone, the angles formed by the neck with the femoral shaft must be known. This presupposes that the axes and the planes used to determine the angles have been defined. Fig. 1 shows the axes and the planes according to Billing and Backman. Their definitions are used in this study. The OH line represents the cervical axis. This axis

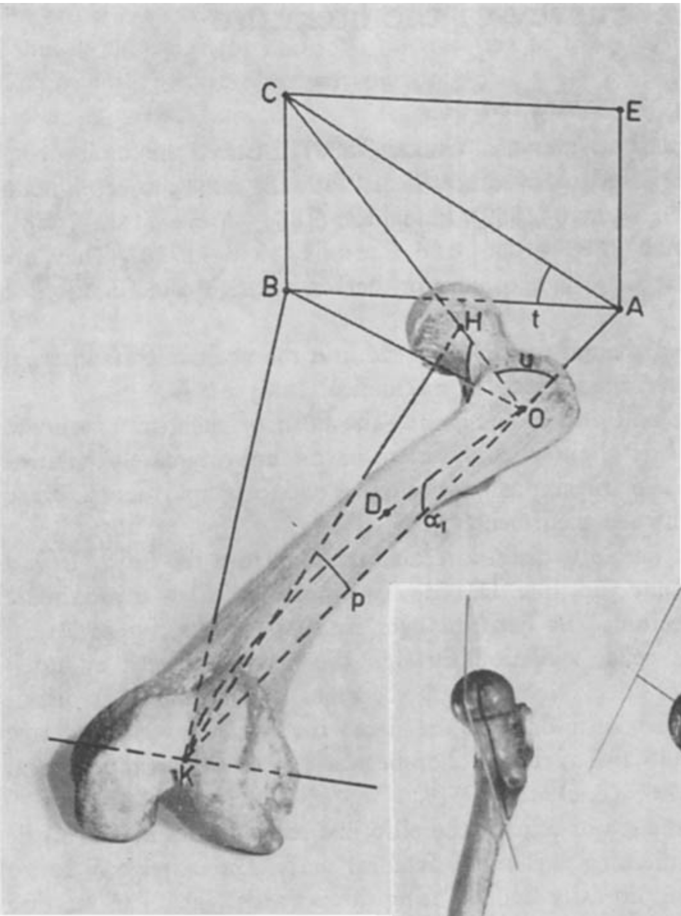

Fig. 1. *The geometrical skeleton of the femur, illustrating applied angles and planes. Backman (1957).*

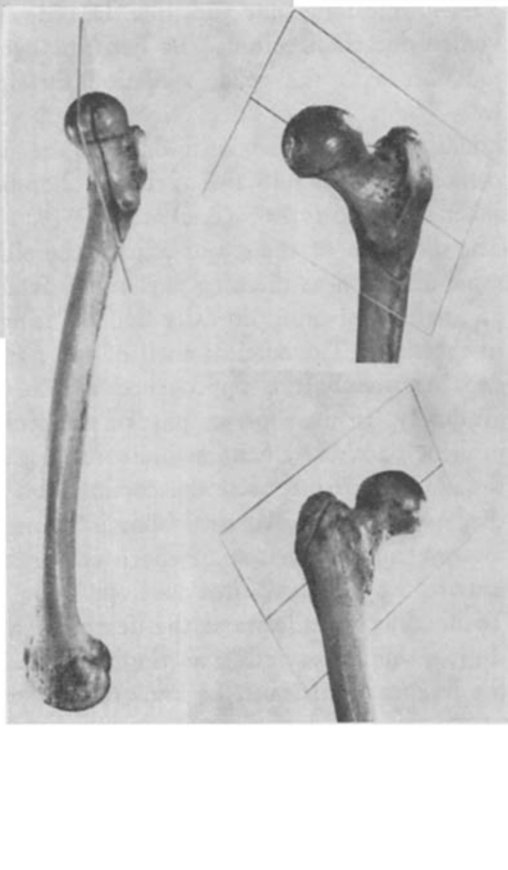

Fig. 2. *The principle plane of the femoral head and neck. Backman (1957).*

intersects the centre of the head H and at the point O the longitudinal axis ODK of the diaphysis. The axis ODK, similar to the femur, forms a ventro-convex curve. As this is unsatisfactory in geometrical calculations, Billing (1954) introduced what he called the ideal axis OK, which is a straight line, connecting point O with point K where the axis of the diaphysis perpendiculary intersects the line between the femoral condyles. Using the above-mentioned reference points, two principal planes perpendicular to each other can be defined; the frontal plane of the femur which coincides with the ideal axis and the condyle axis (plane ABK) and the sagittal plane of the femur, which coincides with the axis of the diaphysis and the ideal axis (plane AEK). Their line of intersection, line AOK, together with the cervical axis OHC form a geometric system.

The angle, formed by the neck of the femur and the shaft, the cervico-diaphyseal angle, is defined as the angle formed between the cervical axis OHC and the ideal axis AOK. Its supplementary angle in figure 1 called u is on the average 54°. The cervico-diaphyseal angle can also be defined as the angle formed by the cervical axis OHC and the proximal longitudinal axis of the femoral shaft OD. This angle is slightly smaller than the angle formed by the cervical axis OHC and the ideal axis AOK, depending upon the magnitude and direction of the torsion angle.

The cervical axis OH usually deviates forward to the frontal plane of the femur ABK and forms with this plane an angle v. A plane through the ideal axis and the cervical axis AOKC is called the antetorsion or the anteversion plane. The magnitude of antetorsion is given by the angle between the frontal plane of the femur AOKB and the antetorsion plane AOKC, the angle t. This angle is on the average 14°, but the variation is great (Backman 1957).

If a plane is laid through the head and the neck of the femur coincident with the cervical axis and the major axis of the section area (fig. 2), this plane will run through the middle of the fovea and bisect the head and the neck into two symmetrical halves. It intersects the femoral shaft medially and distally just ahead of the lesser trochanter. This plane, called the principal plane, is coincident with the major axis of the section area, and forms a dorsally open angle ω_1, pointing downwards, with the cervico-diaphyseal plane, which is on the average 25° and with the antetorsion plane (AKC in figure 1) an angle approximately 15°.

The variations of the cervico-diaphyseal angle, the antetorsion angle and the principal plane under normal and pathological conditions have been the subject of different mechanical interpretations. Opinions are held by some investigators that changes can take place in the design of bone in order to permit yielding to existing mechanical stresses while others claim that changes occur because of yielding. In addition to this, genetic factors also play a part.

In the newborn the cervico-diaphyseal angle is about 150°. At the age of 18 months it usually has diminished to 140° and in adults it is about 130°. In old people the cervico-diaphyseal angle may be noted to have been reduced to about 120°, (Lanz & Wachsmuth, 1938). Humphrey (1888) and Walmsley (1914—1915) maintained that the cervico-diaphyseal angle will be larger if the femur is not subjected to load during growth. Storck (1943) maintained that in paresis of the abductors, the centre of gravity will be centered over the head of the femur in gait and this would mean a more vertical direction of force. In order to resist this vertical direction of force the neck of the femur will turn to a valgus position.

The ventral deviation of the neck beginning early in foetal life and increasing until delivery, probably depends on a torsion of the upper femur. Consequently, the term version is less appropriate. At delivery the antetorsion is about 35°. After delivery the torsion decreases and at the end of bone growth the anteversion has reached its final value, usually around 14°. The variation, however, is large and Backman (1957) in his series found the largest angle at 33° and the smallest at —13°. Weszycki (1957) found a correlation between the degree of antetorsion, the direction of the acetabulum and the size of the cervico-diaphyseal angle. In coxa vara and a steep acetabulum there is slight antetorsion, while in coxa valga the acetabulum is more horizontal with a large antetorsion angle. The fact that the antetorsion increases during the uterine life to become reduced after delivery has been the subject of both genetic and mechanical interpretations. Le Damany (1903) believed that antetorsion is due to reduced space in the uterus and thus an adaptation to the squatting position of the foetus. The author has based his assumption on experimental studies on animals. Graf (1909) who made anatomical studies on the foetus doubts this explanation. Graf pointed out that twins have no increased antetorsion, which would serve as a proof against Le Damany's theory. Friedländer (1909), Brandt (1928) and Backman (1957) believe that the antetorsion depends on rotation in the femur during the foetal life. Grünewald (1912) and Lange and Pitzen (1921) believe that the deviation is a combination of version and torsion. Lange and Pitzen also hold the opinion that the cause of antetorsion as suggested by Le Damany ought to give retrotorsion instead. The anteversion reduction during the growth period according to Morscher (1961) may also be due to the action of the internal rotators in gait. Frankel (1960) utilizing load tests with vertical force direction found larger compressive stress in the posterior portion of the neck of the femur than in the anterior.

Even if mechanical interpretations are justified, no doubt the development is influenced also by genetic factors. Martin (1959) has shown that different races have different antetorsion values. Hultcrantz (Steindler 1955) found that races who usally sit in a squatting position have an increased antetorsion. Altman (1924), Dega (1933) and Badgley (1943) believe that antetorsion is a genetic adaptation to the erect position and according to Dega, antetorsion gives increased stability in a standing position.

The elliptical shape of the neck develops during adolescence. Backman (1957) found that in small children the neck of the femur is cylindrical. Before adult life the final shape and inclination of the neck are formed. Whether this is a genetic phenomenon or an adaptation to mechanical stress or a combination of both, has not yet been proved. Bourgery (1832) and Wyman (1850) wrote that the increasing elliptical shape of the neck in a distal direction strengthens the resistance against vertical load. Ward (1838) drew attention to the elliptical shape of the neck and to the fact that the major axis forms an open angle dorsally. Lehmann—Nitsche (1895) and Strasser (1917) observed the inclination of the major axis of the elliptical cross section, and Lehmann—Nitsche tried to measure the angle on specimens. Grünewald (1918—1919) believed that the inclination of the principal axis was caused by muscular action. Backman (1957) concluded that the inclination of the principal plane means that the force acting on the hip-joint must be applied to the head from the ventral side. This questions the direction of the force that Pauwels (1935) suggested. The fact that the cortical bone is thickest on the inferior side of the neck in the principal plane supports this assumption.

The internal structure of the proximal femur has also been the subject of mechanical analyses. The spongiosa forms a trabecular network in which three main systems can be separated, the medial, the lateral and the arcuate system.

The purpose of the trabecular system is still discussed. Usually the medial part is, as Bourgery (1832) stated, thought to take compressive strains. The function of the arcuate portion is unclear, but it is often believed to take up tensile strains. Storck (1943, 1947) Farkas et al (1948) and Garden (1961),however, believed that there are mainly compressive forces affecting the upper femur. Hirsch and Brodetti (1956) have tried to measure the amount of force carried by the trabecular system and found that the spongiosa was responsible for 30 per cent of the resistance in the femoral neck. Bourgery (1832) described the internal architecture of the femoral head and neck and claimed that trebecular system increases the strength of the bone. He felt that the medial trabecular system carries the load acting on the head through the neck to the cortical layer of the femoral shaft. He also described the two triangles formed between the different trabecular systems; the medial system later became known as Ward's triangle.

Among others Langerhaus (1874), von Friedländer (1904), Farkas et al (1948), Tobin (1955), Rossi (1963 a) and da Silva (1964) have described and analysed the internal architecture of the upper femur. Zschokke (1892), Walkhoff (1904) and Smyth (1958) pointed out that the trabecular system is found in some mammals. Zschokke examined the trabecular system in horses and Walkhoff showed that the medial trabecular system is found in monkeys, although not so fully developed as in man. He also analysed the architecture of the coxal femur end in the Neanderthal man and found that this race no doubt walked in an upright position. The Neanderthal man lacked Ward's triangle. The figures show that the principal axis of the neck is slightly inclined also in the Neanderthal man. Smyth pointed out that animals who spend their lives hanging in trees lack the trabecular system in the hipjoint. Zschokke, von Friedländer and Weidenreich observed that during uterine life there is a vascular network present in the bones whose structure, resembles that of the later developed spongiosa.

Ward (1838), Wyman (1850) and Humphrey (1858) advocated the theory that the shape of the trabecular system depends on mechanical stress and that the arrangement of the spongy bone has been adequately adapted to these stresses. Ward compared the upper femur end to a lamp bracket. The triangular space formed in the frontal plane bears Ward's name. Humphrey (1858) pointed out that the trabecular system may disappear in older people due to resorption.

At a congress in Zürich in 1866, Meyer introduced a paper published in 1867 on the architecture of the spongiosa. Similar to Ward, he described the direction of the spongious lamellae and the triangle. Among the audience was Culmann who, dealing with graphostatics, pointed out that the structure of the lamellae and their direction were similar to the stress trajectories in a loaded Fairbairn crane. Thus the Culmann—Meyer crane theory was developed, giving rise to both critical and favourable remarks. Wolff (1870—1900) in agreement with the Culmann—Meyer theory established the so-called Wolff's law. Wolff believed, similar to Culmann, that the load on the head of the femur is approximately 30 kg. Wolff based his theories partly on studies of pathological conditions. A vital point in the Culmann—Meyer crane-theory was believed to be the magnitude of angles under which the two trabecular systems cross each other. If the trabecular systems really represent stress-lines, they should cross each other under an angle of 90°. Prior to Meyer's paper, Ward (1838) discussed the angle under which the bone lamellae are crossed. Wolff and later Koch (1917), Pauwels (1948) and Rossi (1963) held the opinion that the angles almost were 90°. Jansen (1920) on the other hand believed that the lamellae were not perpendicular and questioned whether pressure and tension were the only formative factors. Pauwels (1948) was of the opinion that the trabecular pathways, even if they are not stress lines, run in the same direction as the stress lines. In his paper on functional adaption (1893—1896) Roux partly agreed with Wolff's law and introduced the expression "functional adaptation". Ritter (1888), Lorenz (1893), Bähr (1900), Ghillini et al (1902) and Küntscher (1935, 1936) delivered criticism against the Culmann—Meyer crane-theory. The criticism is based on the load conditions not being as simple as had been assumed by Meyer. The shape of the bone is not entirely dependent on mechanical factors, therefore genetic conditions must be considered. Ritter, inter alia, pointed out that it is necessary to know how the femur is loaded in order to decide whether the lamellae run in accordance with the stress lines. Triepel (1904—1922) suggested that 1) formation of the spongiosa is an expression of a "harmonious adaptation", and 2) formation of the spongiosa is modified by the stress to which the bone is subjected. The author dismissed Roux' tests on models (Roux 1893) as he believed that these tests could be valuable only if the magnitude of the stress was known. If the internal architecture is a sign of stress during load, all forces including those of muscles must affect the orientation of the bone lamellae. According to Grünewald (1912) the musculature is of greater importance in forming the femur than the body weight. The shape of the bone is therefore also dependent on the force created by separate muscles as they are applied to the bone. Hanausek (1914) discussed the architecture of bones in relation to the compressive and tensile stresses which appear and Koch (1917) tried mathematically to analyse the architecture in order to determine the quantitative relations existing between structure and function. Koch believed that good adaptation exists between the internal structure and the mechanical demands for each part of the bone. The internal architecture of the bone gives a maximum strength with a minimum of material. Haughton (1864), Jores (1920), Carey (1929), Murray (1936), Milch (1940) and Knese (1955, 1956 and 1957) pointed out that the muscular forces probably exceed the static load, and that the dynamic stress is of the greatest importance to the architecture of the spongy bone. This opinion is also held by Jores and Murray who claimed that intermittent pressure stimulates the growth of bones. According to Carey the trabeculae are arranged along the stress lines and arise through muscular activity. The author also believed that the calcar femorale is caused by stress from the hip flexors, the ileopsoas muscle and the gluteus maximus. Milch maintained that the direction of the resulting force can be determined by analysis of the internal structure. Gluckmann (1941—1942) believed that tensile stress provokes the formation of bone and affects the architecture. Pauwels (1948, 1952, 1955 and 1958) on the basis of theoretical analyses and experiments believed that the trabecular system is built up in order to achieve a framework with minimum of material. Bone apposition and resorption are caused by the magnitude of stress and, due to remoulding, the lamellae will have the same direction as though they were stress lines. This means that the direction of the lamellae coincides with the pressure directed toward them. If the stress conditions are altered, a new orientation will result. Stress on the epiphyseal cartilage may, according to Pauwels (1952, 1958) affect the bone growth. Tensile stress on the epiphyseal cartilage would stimulate growth in length, while thickness is determined by compressive stress. Kummer (1955—1959) agreed with the theory that the skeleton is built up with a minimum of material in order to resist the required stress. Furthermore, the upper femur can be regarded as a crane, but he disapproves of photo-elastic tests on models for analysis of the trabecular system, as the magnitude and direction of the forces acting on the hip-joint are unknown. Smyth (1958, 1964) inquired about a quantitative method for measuring forces in the hip-joint, as the muscular force is difficult to evaluate. The author believed that bones can be formed to resist stress in the best possible way. Garden (1961), on the other hand, believed that the internal architecture of the femur cannot be determined by mechanical principles, but that the structure arises through rotation and expansion from the shaft of the femur. The spongy architecture depends on the expanding and rotating lamellae.

Forces acting on the hip-joint.

The hip-joint is surrounded by large muscles and strong ligaments. The muscles and, to a certain extent, the ligaments affect and complicate the calculations of the forces which act on the head of the femur. Lack of knowledge about the magnitude and direction of these forces under differing circumstances has rendered the understanding of the biomechanics of the hip-joint difficult. In their analyses, Meyer and Culmann assumed a load on the femoral head of about 30 kg. Criticism has been directed against that low load because the authors were believed to have disregarded muscular effect. However, Meyer was aware that the force acting on the head of the femur is influenced by muscles. Meyer (1849) pointed out that the joint cartilage is subjected to a pressure which is determined both by gravity and muscular force. Fick (1850, 1903), Milch (1940) and Debrunner (1958) pointed out the importance of muscular action on the hip-joint.

Many attempts have been made to calculate the resultant force which affects the head of the femur with regard to both size and direction.

Henke (1868, 1869), Thomsen (1934) and Küntscher (1936) pointed out that when calculating the force, both gravity and muscular forces must be considered.

Pauwels (1949, 1950—1951) pointed out that the centre of gravity lies eccentrically in relation to the longitudinal axes of the bones and this may provoke heavy bending moments. Muscles and ligaments increase the compressive forces but reduce the tensile stresses. The long bones are composed of a minimum material.

Koch (1917) calculated the dynamic load at double that of the static. In gait the static load on the head of the femur is about 80 per cent of the body weight and the forces acting against the head of the femur will be 1.6 times the body weight. Koch pointed out, however, that the femur may act as a shock absorber due to flexion in the hip and the knee. Grünewald (1918) believed that due to muscular action, the force acting on the hip-joint may reach 400 kg. Storck (1931) tried to calculate the force which acts on the hip-joint, with respect to both magnitude and direction. The muscular force was calculated at double the body weight. The distance between the line of action of the centre of gravity and the centre of the head was calculated at 10 cm and the distance between the line of action of the musculature and the centre, 5 cm. When standing on one leg this would mean a force of about 200 kg. Pauwels (1935) on the basis of Fischer's works has calculated the magnitude and direction of the force acting on the head of the femur in one-leg support, two-leg support and during the stance-phase in gait. During one- and two-leg support insignificant muscular forces act in the horizontal plane and therefore the horizontal component of force is considered negligible. In gait a dynamic component is added in the horizontal plane, but the dynamic force in the frontal plane is predominant. The author believes that insignificant muscular forces act when standing on two legs and the vertically directed force is equal to about $^{1}/_{3}$ of the total body weight. In one-leg support, the muscular forces affect the magnitude and direction of the resultant. The centre of gravity for the superimposed body weight is assumed to be 10.9 cm from the centre of the head, and the lever of the abductor muscle was estimated to be 40 mm. The force acting on the head of the femur will be about three times the total body weight. If, during the stance-phase, consideration is given to the dynamic forces, the force will be $4^1/_2$ times the body weight. The horizontal component, which is dependent on the speed of gait, reaches its maximum at the beginning of the stance-phase and amounts to about 73 per cent of the body weight. It is then reduced to a minimum, and at the end of the stance-phase it again reaches a maximum of about 38 per cent of the body weight. When using a walking stick the vertical force can be reduced to approximately 1.7 times the body weight.

Inman (1947) calculated the force which acts on the head and believed that while standing on one leg, the force which acts against the head of the femur is 2.4—2.6 times the body weight.

According to Cabot and Peralba (1952) the force on the femoral head in gait is equal to the fourfold body weight plus half of the body weight for a dynamic effect. Knese (1955) believed that the pelvic resultant due to muscular forces in a superimposed body weight of 57 kg will be 395 kg which means that the force will be about 4.6 times the body weight. Müller (1957) has calculated the force on the head of the femur in gait at 4.5 times the body weight, for a person weighing 75 kg, which would mean a load of more than 330 kg. The author further pointed out that the compressive stress in the hip-joint is 3 times as great as the tensile stress, depending on the direction of the resultant. In two-leg support, the force, according to Denham (1959), which acts on the hip-joint is slightly greater than superimposed body weight and this is dependent on the muscular activity required to maintain equilibrium. In standing position on one leg and a body weight of 68 kg, the joint will carry a vertical load of 204 kg and the vertical component of the abductor muscle force will be 136 kg. If the neck of the femur is shortened, even larger forces would occur and, therefore, in cases of arthroplastics the length of the neck should be maintained when possible. If, on the other hand, the head of the femur protrudes into the acetabulum the force will be reduced. Abductor tenotomy would, according to Denham, increase the load on the hip-joint. A medial displacement of the femoral shaft would not alter the force conditions. For instance, walking with a stick may reduce the force from 204 kg to 54 kg. Osborne & Fahrni (1950) and Soren (1963) also pointed out that by altering the levers, the force acting on the hip-joint can be changed. Hackenbroch (1961) found that when standing on one leg the force on the femoral head is equal to the fourfold body weight minus the weight of the supporting leg. During gait the resultant will correspond to the fourfold body weight due to dynamic forces. Williams and Lissner (1962) calculated the force acting on the hip-joint in standing position on one leg, at 2.4 times the body weight. Hochman (1964) found that the force acting on the hip-joint may amount to six times the body weight. Rossi (1963 b, c) believed that when standing on two legs a force of 25.1 kg acts on the hip-joint at a body weight of 80 kg. While standing on one leg utilizing the same body weight, the static force will be 276.1 kg and the muscular force 211 kg. While walking, the forces acting on the hip-joint are not as great as is usually assumed according to Rossi, depending on the displacement of the centre of gravity during the stance-phase. The author described the variation of the force during the stance-phase and obtained a maximum value of 148.8 kg at a body weight of 80 kg. Hauge (1965) pointed out that while standing on two legs the load will be half the overlying body weight plus a force of unknown magnitude. This additional force depends on the muscles required to maintain equilibrium. When standing on one leg the author calculated the load on the hip-joint at 180 kg if the body weight was 60 kg.

Due to the muscles the forces resultant will not be quite vertical. Most previous calculations have been made only for the frontal plane. Storck (1931) believed that the force in the standing position on one leg forms an angle of 15° with the vertical plane. Pauwels (1935) believed that standing on two legs gives an entirely vertical force because the muscular components are insignificant. Standing on one leg the force in the frontal plane forms a 16° angle with the vertical line. Pauwels has also determined the direction of the horizontal component and found that in the horizontal plane it forms a dorsally open angle with the frontal plane of about 30° in the beginning of the stance-phase and at the end of this phase it reaches a minimum angle with the frontal plane of about 2°. This means that the femoral head is subjected to a load on its dorsal side. Inman (1947) believed that the direction of force in one-leg support is constant, irrespective of the position of the pelvis. The force direction forms an angle of 10°—15° with the vertical line in the frontal plane which corresponds to the direction of the medial trabecular system. The epiphyseal line runs perpendicular to the medial lamellae. Since the resultant seems to follow these trabeculae, no consequent shearing force will arise on the epiphyseal cartilage in the frontal plane. During growth the abductor muscles alter their direction of action. The direction of the resultant alters as well and the epiphyseal plate rotates so that it will always lie perpendicular to the direction of force. When the abductor musculature becomes paralysed, the load will be more vertical. This is one reason why the epiphyseal cartilage remains in its horizontal position, and gives valgus position. Knese (1955) assumed that the pelvic resultant affects the head of the femur obliquely from above and then runs into a ventro-caudal-fibular direction. Backman (1957) believed that the force in one leg-support forms an angle of 28° with the longitudinal axis of the neck. This direction coincides with the direction of the medial trabecular system.

The magnitude of the pressure arising on the head of the femur depends on the size of the contact surface. It is difficult to assess this exactly, especially since one cannot be certain that the centre of the contact surface coincides with the point of force application. According to Fick (1904) the medial upper portion of the head of the femur is subjected to the greatest pressure since its lateral upper portion lies outside acetabulum. The author found this to be in accordance with Werner's (1897) calculations on the thickness of cartilage. Little, Lionel, Trueta (1948), Harryson, Schajowiz and Trueta (1953) and Trueta (1954) classified the head of the femur as containing a pressure-area and a non-pressure-area. Buchet (1959) and Castaing et al (1960) pointed out that the anterior part of the femoral head is of great importance to the mechanics of the hip-joint, especially when the anteversion angle is large. Together with Chevallier and Plisson (1963) Castaing reported a weak zone to be situated in the upper anterior part of the hip-joint. While standing, the resultant runs along the longitudinal axis of the neck and forces the head of the femur upwards, inwards and forwards. The maximum load will, according to the authors, be carried by the anterior upper quadrant of the acetabulum.

Stress and strain in upper femur.

Many attempts have been made to determine stresses and strains in the femoral neck, when loaded. Merkel (1874) pointed out that the proximal femur must carry a heavy load under an unfavourable angle. Merkel believed that analyses usually are applied to one plane only, and this is a disadvantage. Due to the anteversion of the femoral neck the force acting on the head will attempt to break the neck of the femur in the ventral direction and not in the longitudinal direction as Meyer and Wolff have assumed. This means that the most stressed areas are the posterior and anterior portions of the femoral neck.

Attempts have been made to quantitate stress by means of the stress coat technique, photo-elastic model studies and strain-gauges. As bone is a heterogeneous substance the relation between stress and strain is rather complicated. Many investigators are aware that these tests give only limited information.

Roux (1895) carried out load tests on a rubber model which had been treated with paraffin. Triepel (1904) criticized Roux' investigations and mentioned that no conclusions can be drawn when the force direction is unknown.

Küntscher (1935 a, b) carried out stress coat investigations on femurs which were subjected to dynamic loading under different inclinations of the femoral shaft. The same stress was obtained by different experiments. The stress lines had a different course on the front of the femur as compared to the back, a phenomenon ascribed to the presence of the calcar femorale (1936). The author used a mirror extensometer for the measurement of bone deformities. He pointed out that due to the heterogeneous nature of the bone, stress could not be mathematically calculated. Milch (1940) carried out stress analyses by means of models manufactured from thin Catalin plates. The experiments were carried out in two dimensions, but the author believed that three dimensional experiments would be of greater value. During the experiments, stresslines were obtained corresponding to various trabecular systems by loading the model in abduction- and adduction positions. The author pointed out the difficulty of assessing the muscular forces in the hip-joint since the magnitude varies from time to time. Pauwels (1948, 1950, 1951 and 1955) carried out photo-elastic model experiments and the results obtained showed that the architecture of the bone is adapted to the stress to which the bone is subjected, and that bone is formed from a minimum of material. Pauwels furthermore believed that the muscles serve to eliminate the bending moment. The musculature is required to develop its entire force, when needed to equalize compressive and tensile stress. Evans and co-workers have made numerous strain analyses in the upper part of the femur. Stresscoat studies have been performed using a vertical load and with the intracondylar plane horizontal or with a laterally open angle of 3° (Evans 1948, Pedersen, Evans and Lissner 1949), Evans, Pedersen and Lissner 1951, Lissner and Evans 1956, and Evans 1962).

Rauber (1876), Koch (1917) and Knese (1956, 1957), pointed out the impaired resistance to tensile stress of bone and this has been verified by Evans and co-workers. Furthermore it was noted that the speed used to apply the load is of importance to the bone's resistance. This has also been proved by Mc Elhaney (1965) and Sedlin and Hirsch (1965). In a paper by Evans and Charles (1957) dealing with stress-coat technique, the so-called splitline method was used. According to Evans and King (1961) spongy bone shows a poor resistance to load, but has good energy-absorbing capacity. Jakobsson (1954) carried out photo-elastic model tests where stress-distribution in normal and dysplastic hip-joints were studied. Kummer (1955—1956, 1959) carried out stress-coat experiments on plexiglass models. Fessler (1957) found in photo-elastic studies on a hip-joint model, stress lines that resembled the trabecular system. These experiments were carried out on one plane only. Hirsch and Frankel (1960, 1961) performed analyses with the strain-gauge and the stress-coat techniques. The abductor muscles were found to cause compression in the upper portion of the neck and the distribution of stress in the femoral neck was affected.

Measurement of dynamic forces during gait.

Determination of forces acting on the femoral head in walking is often combined with gait analyses. It is often desirable to correlate the forces to the different phases of double steps.

For a detailed history on gait analyses, reference is made to Weber (1836) and Steindler (1955). Besides static load and muscular forces, dynamic forces are also involved in gait. These are determined by the body mass and its acceleration. Therefore, gait may cause a load on the hip-joint which considerably exceeds the body weight. The vertical force arising between the foot and the ground may be calculated, but is complicated. Fischer (1899-1904) determined this mathematically.

Pauwels (1935) on the basis of Fischer's data calculated both the vertical and the horizontal dynamic force acting on the hip-joint.

An apparatus constructed to record the forces developed between the foot and the ground will facilitate the measurements. The vertical force which acts on the hip-joint should in relation to time coincide during the stance-phase with the vertical force between foot and ground.

The first attempts to determine the force between foot and ground in gait were made by Carlet (1872) and Marey (1873, a and b, 1883, 1885, 1894). Queneu and Demeny (1888) performed similar measurements. The measuring apparatus was called a dynamograph and consisted of two air-filled pockets, one placed under the heel and one under the fore-foot. By means of pressure variations the vertical force component between foot and base was determined as well as the time of loading. The diagrams of the anterior and posterior pockets were drawn. The graph obtained is the same shape as those obtained when recording with the present electronic force-plates. From the graphs the various phases of gait could be determined. During measurement the vertical force did not exceed the body weight by more than 20 kg even in rapid gait. Measurements were performed together with chronophotography. In some cases, the vertical force was less than the body weight during the stance-phase. The author believed that those cases, where pressure exceeds body weight, are dependent on the vertical acceleration of the centre of gravity. Schwartz and Vaeth (1928) constructed a so-called basograph for measuring forces between foot and base. The design was abandoned and Schwartz and Heath (1932) introduced a pneumograph for recording these forces. The design was made of pneumatic cells, fitted in the heels of specially manufactured shoes. They were connected with a recording instrument and variations in pressure between foot and base were determined. By letting the subjects walk on a controlled speed tread-mill, the walking speed could be determined. Film recording was performed simultaneously. In 1934 the pneumatographic design was dispensed with and replaced by an electrographic recorder which recorded both the total time of loading and contact time of the heel, 5th metatarsal head and the great toe (Schwartz et al 1934, 1936, 1937, 1941, 1949 and 1964). Elftman (1938—1939) improved the technique and built a mechanical force plate which served to determine three force components between foot and ground. Measurements were carried out with simultaneous photography. Eberhart (1947) published a report from the University of California which dealt with studies on gait. Measurements were made with electronic force plates. Early studies used a force plate, constructed by Northrop Aircraft Corp. but later Eberhart's own design was employed. The forces arising between foot and ground are obtained in electronic force plates by means of strain-gauges. All force components can be recorded. Eberhart and Inman (1949), Elftman (1949), Cunningham and Brown (1952), Saunders, Inman and Eberhart (1953), Cunningham (1958) reported measurements with force plates.

For gait studies of horses, Björk (1958) designed a force plate mounted in a horseshoe, which determined the force arising between the ground and hoof under different conditions. Wetzenstein using a similar technique, (1964) mounted a power plate in the heel of a man's shoe and determined the load conditions of the heel in gait.

Marks and Hirschberg (1958) and Drillis (1958) used force plates to analyse the forces between the foot and ground under normal and pathological conditions. In order to study friction on different floor surfaces by means of force plates, measurements were performed, inter alia by Harper, Warlow and Clarke (1961) and Carlsöö (1962). The first-mentioned authors have, in some cases, recorded a vertical load less than body weight during the entire stance phase. Contini (1964) worked with electronic force plates and believed that in rapid gait the vertical force has a certain maximum value which is not exceeded.

II. Design of the measuring prosthesis

Introduction

Ingelmark and Blomgren (1948) have designed an apparatus for pressure-measurements in joints under vital conditions for experiments in animals. No attempts have been made to measure forces in the hip-joints of human beings in vivo. In the literature chapter earlier attempts to determine the magnitude and direction of the forces acting on the femoral head under static conditions have been reported. Attempts to calculate the dynamic components have been made by Pauwels (1935). One drawback with theoretical analyses is that they must be carried out for a certain position in the hip-joint, for example, standing on one leg or with a known degree of flexion in the hip-joint. During the standing position on one leg, the line of action of the body-weight does not pass through the centre of the head of the femur, the upper part of the femur is subjected to moments, and to maintain the balance the muscles must exert a counter force on the joint. From the two forces — the weight and the muscular force — the force acting on the head of the femur is then determined by applying the parallelogram of forces. This resultant is then resolved into two perpendicular components. In flexion the position of the centre of gravity in the leg and the direction of the muscular pull must be known if the force is to be determined.

Although under certain circumstances the force acting on the hip-joint can be determined to a fair level of accuracy by this method, it has certain shortcomings. For instance, the calculations are usually made for only one plane, that is to say, only the components P_x and P_y in fig. 3 are determined.

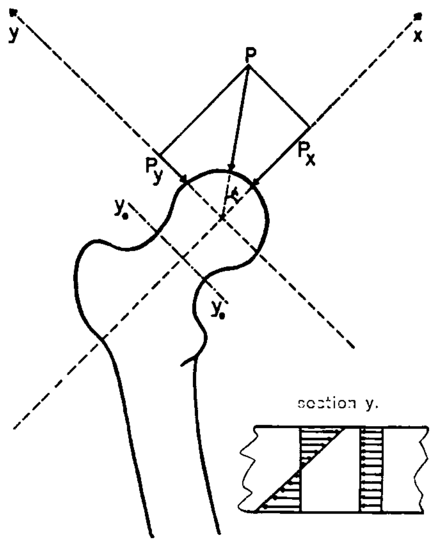

Fig. 3. *A force P acting on the femoral head can be resolved into two perpendicular components P_y and P_x. In a section through the neck the strains caused by the forces are indicated.*

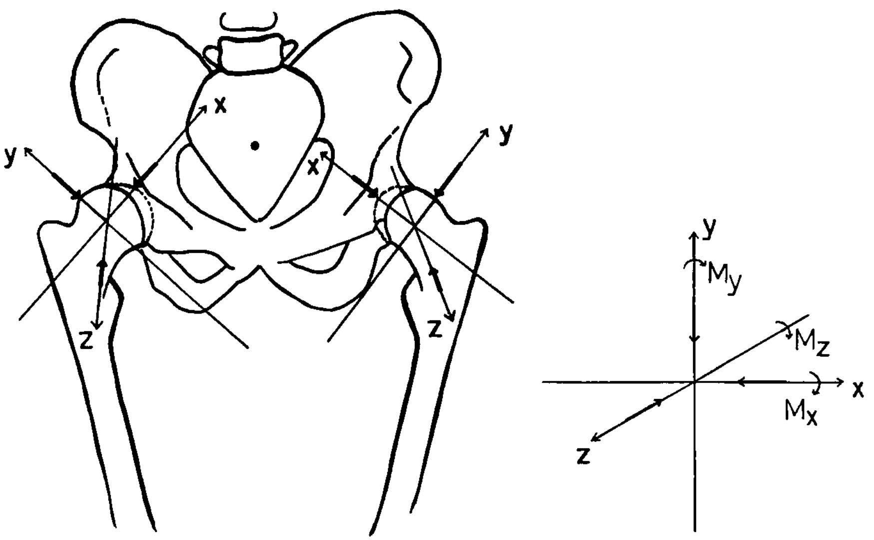

Fig. 4. *A three dimensional co-ordinate system is applied to the femoral head and neck with origin at the center of the head and the x-axis along the longitudinal axis of the neck. The force on the head can be resolved in three orthogonal components acting along the axes. The components are nominated in accordance with the axes they act along, and the moments set up in the neck in accordance with the axes they act about.*

The results will be valid only in the cases in which the force acting in the third plane — the horizontal component — is negligible. If the values are to be accurate it is necessary to perform the calculations for certain situations — for example with the subject standing on one leg, when the horizontal component is small and the force acting is large. By adding a value for the dynamic effect, the force acting during the stance-phase in walking would be obtained.

Though one-leg support is a position in which the upper end of the femur is subjected to the greatest static force, there are others more common in which it is of interest to know the forces acting on the head of the femur. In, for instance, flexion of the hip-joint the components P_x, P_y and the horizontal component P_z should be more equal in magnitude, and then the application of the parallelogram of forces in one plane will not give very accurate values.

In flexion and extension or abduction and adduction it is difficult to determine exactly the direction in which the muscles pull and the length of the moment arm, and this introduces a factor of unreliability in the calculation of the forces acting on the hip-joint.

More reliable results would be obtained if the force acting on the femoral head is determined in relation to a three dimensional co-ordinate system (fig. 4). If three orthogonal components are to be measured simultaneously, they are measured along the x-, y- and z-axes, and the forces acting along these axes are called P_x, P_y and P_z respectively. If the co-ordinates are placed as in fig. 4, with origin at the center of the femoral head and the x-axis along the longitudinal axis of the neck, P_x will cause compressive strains in the neck, while P_y and P_z will cause bending in the neck. The bending moment resulting from P_y occurs about an axis parallel to the z-axis, and that from P_z about an axis parallel to the y-axis. Accordingly, these moments are called M_z and M_y respectively.

Forces acting on the hip-joint are usually not static; instead there is, from the mechanical aspect, a relatively slow dynamic development, the point at which the force acts on the head of the femur changing its position as the head moves in the joint. Since one component of the force P acting on the hip-joint is the muscular force, the direction at which P acts on the head alters as the end of the femur moves in the acetabulum. It is almost impossible to calculate the motion of the point of action of the resultant force over the surface of the head in, for example, the different stages of walking — the swing-phase and the stance-phase.

If it were possible to measure these forces directly on the living subject more reliable information would be obtained and the dynamic course of events could be recorded.

The force acting on the head of the femur causes stresses and strains in the neck of the femur (fig. 3). In theory, therefore, it would be possible to determine the stress by means of strain gauges applied to the neck of the femur, and from the recorded values to calculate the components of the force acting on the head. In practice, however, this is not feasible; for problems arise in applying strain gauges under vital conditions; for instance, it is difficult to attach the gauges stably to the moist and slightly greasy surface of the bone. The conducting material must be insulated against moisture; and measures must be taken to avoid any toxic effect of the cement.

Since the physical properties of the bone differ not only from one person to another but also intraindividually, according to the local conditions, (Carothers et al 1949, Evans and Lebow 1951, Evans 1957 and 1964, Forsblad 1959, Hirsch and Evans 1965, Sedlin 1965), the measurements are not reliable generally. Bone is a heterogeneous material and Hooke's law is not obeyed completely. Since also the geometry of the neck varies, accurate measuring values cannot be obtained unless calibration can take place by applying a known load, and this is impossible in experiments under vital conditions. This method, however, can be applied to model tests on autopsy specimens (Hirsch and Frankel 1960, 1961, Frankel 1960).

Since metal is a highly suitable material as a support for a strain gauge, and the head and neck of the femur can be replaced by a metal prosthesis, it was considered that the required forces might be determined by placing such gauges in a prosthesis that could be inserted in the upper end of the femur. The results thus obtained are not directly applicable to physiological conditions, but the method was believed to give new information on the hip.

The commercially available hip prostheses are not suitable in their existing form for the construction of a prosthesis of the type desired; for most of them have no neck part, and this is indispensible because the essential feature of the method is that the forces acting on the hip-joint are determined from stresses set up in the neck part. If the measurements are to be accurate the dimensions of the prosthesis must be exact, and the strain gauges, which are not well tolerated by the tissues, must be applied so that they cannot come into contact with the body fluid. The best way of ensuring this is to place the gauges inside the prosthesis.

In the design of a prosthesis to replace the upper end of the femur and to serve as part of the measuring system, account must be taken of the manner in which the signals are to be transmitted from the prosthesis to the recording unit. Telemetry would have been the best and the most elegant method but it was ruled out in the present case on practical and financial grounds. The signals had therefore to be transmitted by wires in the conventional way. Before the design, dimensions and selection of the material for the prosthesis could definitely be settled, certain assumptions had to be analysed and definitions established.

Definitions of forces acting on the measuring prosthesis

In the design of a prosthesis of the type proposed with built-in strain gauges the following theoretical assumptions must be made.

Let P be the force acting on the prosthetic head (fig. 5). This force, unknown in magnitude as well as direction, is to be determined. Force P can be resolved into three mutually perpendicular components P_x, P_y and P_z, parallel to a co-ordinate system that is fixed in relation to the prosthesis. The origin of these co-ordinates is placed at the centre of the head, the x-axis coincides with the longitudinal axis of the neck, the y-axis perpendicular to this in the frontal plane, and the z-axis perpendicular to the other two axes (fig. 5). The advantage of recording the forces to a co-ordinate system which is fixed in relation to the prosthesis instead of in relation to the co-ordinates of the body, is that the direction of the force is obtained in relation to the proximal femur irrespective of its position in the joint.

In the absence of friction the force P would act through the mid-point of the head; when friction is introduced P no longer acts at the mid-point; but the displacement is negligible, and in most cases it can be assumed for the purpose of the calculations that P does act through the mid-point.
The component P_x gives rise to a compressive force in the neck of the prosthesis, and the components P_y and P_z give rise to the moments M_z and M_y about the z- and y-axes, respectively. Even around the x-axis there is a moment; due to friction called M_x. M_z and M_y cause compressive as well as tensile strains in the neck, M_x torsional strains.
By placing measuring receptors at a known distance from the point of action of the components, which is, the mid-point of the head, the strain resulting from the force components can be recorded and the forces deter-

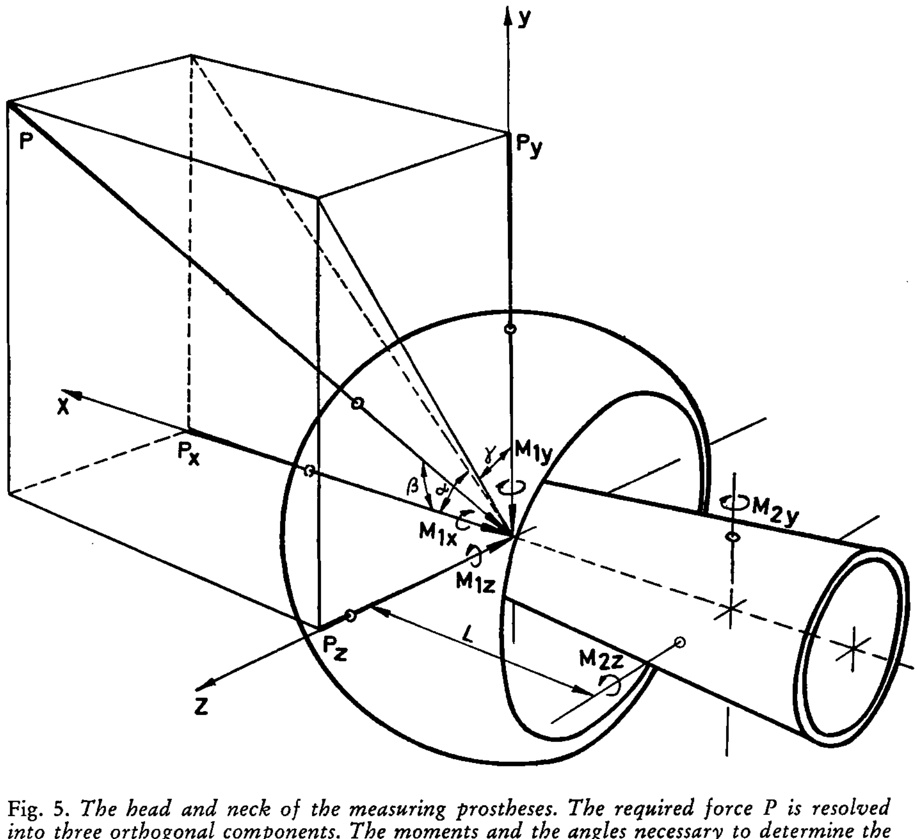

Fig. 5. *The head and neck of the measuring prostheses. The required force P is resolved into three orthogonal components. The moments and the angles necessary to determine the direction of the forces are indicated.*

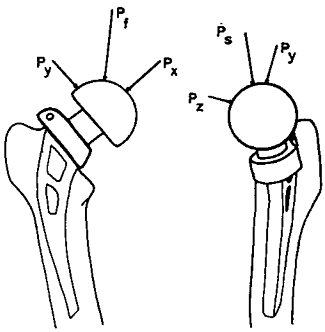

Fig. 6. *The measuring prosthesis applied to the proximal femur with the force components indicated P_f=projection of the force P in the frontal plane, P_s=projection of the force P in the sagital plane.*

mined. One advantage of this procedure is that the components P_x, P_y and P_z are recorded. The magnitude of the resultant P is given by the expression:

$$P = \sqrt{P_x^2 + P_y^2 + P_z^2} \tag{1}$$

The angles under which P acts on the prosthesis are obtained by trigonometry. Two of these angles, α and γ, have been chosen (fig. 7) as representative of the direction of the forces, namely, those formed by the components P_x and P_y and the projection of resultant P in the frontal and sagittal planes P_f and P_s, respectively. These angles are given by the expression:

$$\tan \alpha = \frac{P_y}{P_x} \tag{2}$$

and

$$\tan \gamma = \frac{P_z}{P_y} \tag{3}$$

The angles α and γ give the direction of the force components. The direction of the resultant force P is determined by those angles and the angle β in fig. 5. This angle is formed between the resultant force and its component P_x, and is given by the expression:

$$\cos \beta = \frac{P_x}{P} \tag{4}$$

If forces are required in relation to the co-ordinates of the body, consideration must be given to the position of the upper end of the femur in the acetabulum.

As mentioned earlier, the force P will not act through the mid-point of the head due to friction. If the coefficient of friction is to be determined the combined effect of normal and tangential stresses can in general not be strictly described by a single resultant force. One correct description would be as one force and one moment, both acting off the center of the head and having 3 components each.

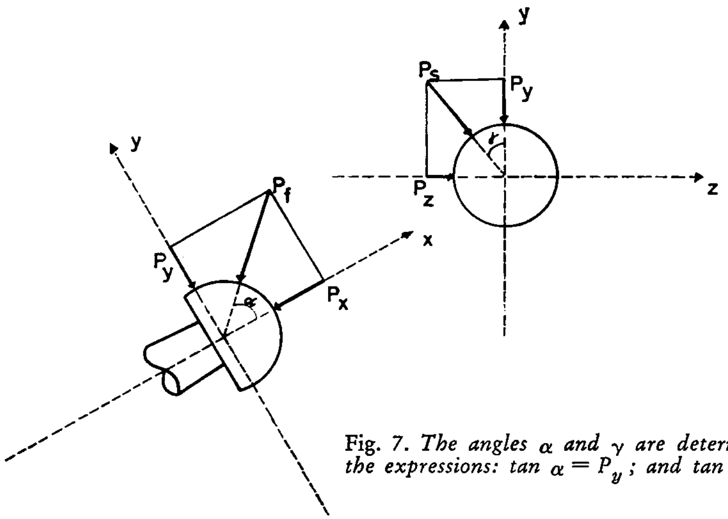

Fig. 7. *The angles* α *and* γ *are determined by the expressions:* $tan\ \alpha = P_y$; *and* $tan\ \gamma = P_z$.

As an alternative way of description, two components of the moment can be explained as the result of a displacement of the resultant force from the center point. This description has the drawback that the displacement can be very large if friction forces are great, which is probably not the case. The stresses due to moments acting in a section through the centre of the head must be recorded if the coefficient of friction is to be determined or if displacement from the midpoint is not negligible for the determination of the force P. Strain gauges have therefore been placed in two neck sections, one section perpendicular to the longitudinal axis of the neck through the mid-point of the head, section I, and the other, section II, at a known distance, L, from section I (fig. 5). The moments recorded in the respective sections are therefore further indicated by the suffixes 1 or 2, such as M_{1y} for section 1 and M_{2y} for section 2. In selecting positive or negative signs for forces and moments some difficulties might occur, since the same force may have different signs for a left and a right hip. The problem has been solved in such way that if the prosthesis is inserted in a right hip and regarded as in fig. 7 the forces acting along the x- and y-axes towards the mid-point are given positive signs. Regarding the z-axis, for practical reasons the force acting along the negative z-axis, that means coming from the ventral side toward the origin, has been defined as positive. If the prosthesis is inserted in a left hip, the signs for the force along the x-axis would be changed, and this is not desirable. Thus, the force along the x-axis is called positive if it acts in medio-lateral direction, along the y-axis the force is positive when acting in cranio-caudal and positive along the z-axis in ventro-dorsal direction. According to the same principle, the signs for moments M_y and M_z are obtained. The torsion moment was recorded only in prosthesis 2 and is positive for rotation in the direction indicated in fig. 5, the prosthesis seen ventrally in the left hip.

If the resultant force P acts through the centre of the prosthetic head, components P_y and P_z are obtained by dividing the moments recorded in section 2, resulting from the components respectively, by the moment arm

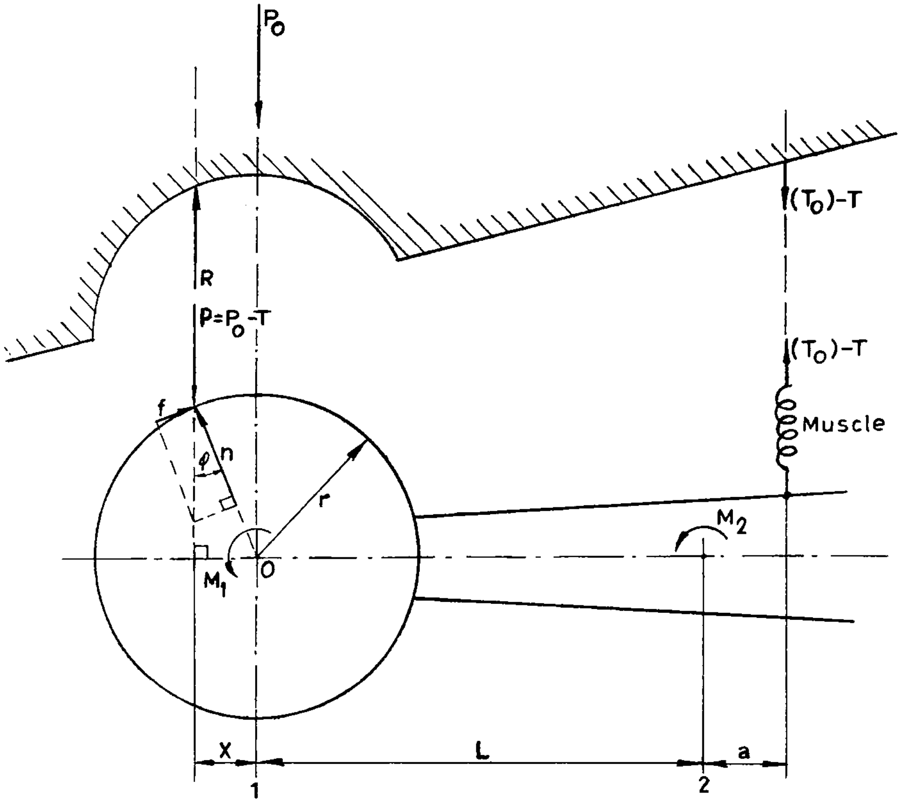

Fig. 8. *Due to friction a force P_0 is displaced from the origin if the muscular force is changed by an amount T. The force of reaction R will result from the normal force n and the friction force f.*

L, that is, the distance between the centre of the head and section 2 (fig. 5). Thus obtaining the expressions:

$$P_y = \frac{M_{2z}}{L} \tag{5}$$

and

$$P_z = \frac{M_{2y}}{L} \tag{6}$$

since the forces are directly proportional to the measured moments. The force component P_x which does not cause any moment is determined directly from the strains resulting from the force.
The above formula does not apply if consideration has to be taken of the displacement from the mid-point of the head of the resultant force P due to friction. Fig. 8 shows how the friction moment can be described as if the resultant is displaced from the origin if the muscular force is changed by an amount T. The force of reaction R, resulting from the normal force n and the friction force f, will also be displaced and balance will be maintained until the friction force is fully established. As long as equilibrium exists the following will apply:

$$P - R = O \tag{7}$$

and

$$T(a+L) - R \cdot x = O \tag{8}$$

Moment section II will give the expression:

$$M_2 = P(x+L) \tag{9}$$

In order to determine P, the friction moment must be subtracted.
Taking the origin as the moment centre will give:

$$M_1 = P \cdot x \tag{10}$$

By subtracting the expression:

$$M_1 = P \cdot x \tag{11}$$

from the expression:

$$M_2 = P(x+L) \tag{12}$$

the following is obtained:

$$P = \frac{M_2 - M_1}{L} \tag{13}$$

Similarly the components are obtained from the expressions:

$$P_y = \frac{M_{2z} - M_{1z}}{L} \tag{14}$$

$$P_z = \frac{M_{2y} - M_{1y}}{L} \tag{15}$$

Photo-elastic experiments

On the basis of previously presented assumptions and theories in order to judge the practical application of the measuring prosthesis, photo-elastic-model studies have been carried out. A model of the proximal femur was made of epoxyresin E. The neck part had the shape of a parallelepiped 18×10×40 mm in size. From these studies information has been obtained about the suitable location of strain gauges. For instance corners radii or fillets will have a disturbing effect and the strain gauges must be placed so that these disturbances will not affect the measuring signals. The neck of the prosthesis must be sufficiently long to permit placing the strain gauges within undisturbed areas. For anatomical reasons, however, the size of the neck is limited.

A cast of Araldite E was made of the upper end of the femur and a measuring section with parallel surfaces was cut out. The stress distribution along the measuring section could be seen when loaded. Information was obtained as to the stress concentration at the fillets and the extent of these stress raising areas. It was thus possible to determine the position of the strain gauges and the desired length of the measuring section. With a 40 mm long neck and a fillet radius of 2.5 mm a highly stressed area of approximately 8 mm in length occurred proximally and distally in the neck. If the strain gauges have an active length of 3 mm, a distance of 20 mm may be allowed between the measuring sections.

Based on data obtained from the photo-elastic tests, a pilot model of light metal, Svenska metallverken 6958, was made. It was provided with strain gauges placed within the limits obtained by means of the photo-elastic tests. The model was subjected to numerous static loads and strains were recorded at the various measuring points. After analysing the obtained data, it was found that a static force could be determined as to magnitude, direction and position with an accuracy of 1—2 %.

Choice of metal

It was desirable to have a neck part which would, when subjected to forces of the magnitude expected, give sufficiently large deformations in the material, so that adequate signals could be obtained from the strain gauges. One condition was also that the material was strong enough to stand actual stresses, in spite of small section thickness.

At present three types of metals are used for orthopaedic implants:

1) Commercially pure titanium. 2) Stainless steel of the 18—8 Mo type. 3) Cobalt-chromium alloys.

Titanium was not used because of its comparatively low resistance to abrasion. A prosthesis of the desired design, made of 18—8 Mo steel will give strong measuring signals, but its strength would be insufficient and the neck might fracture.

A cobalt-chromium alloy is suitable from many aspects; it is strong, inert and has a high resistance to wear (Scales 1965), but is difficult to machine, and was therefore unacceptable owing to the numerous small details in the prosthesis and the need to adhere accurately to the dimensions. 18—8 Mo steel or similar steel such as 17—14—4.5 can be made stronger by strain hardening. This, too, is obtained if a small quantity of titanium is added to a metal alloy and furthermore increases the resistance to corrosion.

The problem has been to find a metal as inert as 18—8 Mo steel and the cobalt-chromium alloys, stronger than 18—8 Mo steel and at the same time having a high resistance to wear. It must be easy to machine. A titanium stabilised stainless steel (Bofors A 286) with the analysis presented in table 1 was selected.

C	0.06 %	V	0.30 %
Cr	15.00 %	Ti	2.00 %
Ni	25.00 %	Fe ca	55 %
Mo	1.25 %		

Table 1. *Metal alloy used for the measuring prostheses.*

The material has a yield strength of 65 kp/mm^2, ultimate tensile strength of 105 kp/mm^2, a hardness of 290 HB and minimal fatigue strength $\pm$ 25 kp/mm^2.

Choice of other material

The wires which are to connect the strain gauges with the recording unit will remain for a long time in the body. As toxic materials are found in wires, gauges and adhesives, it is important that these are isolated from body fluids. As far as gauges and adhesives are concerned, the problem has been solved by placing the strain gauges on the inner surface of the neck part of the prosthesis, which is insulated from its environment. In the leads there is material which might have a toxic effect and therefore each of them has an outer covering of Terylene and together they form a cable encased in a sleeve of Teflon. The leads, Copper Welded 2 × Terylene, are 22 cm long and connected to a multi-pole female contact. This female contact contains toxic material, but this was protected from body fluid by means of a hermetically-sealed metal cylinder made of the same material as the prosthesis.

The Teflon cable and the leads must be removed when the measurements are finished. The Teflon cable could be pulled out and the leads were to be cut off close to the prosthesis and removed. Any part of the leads that remain would come into contact with body fluid, but due to the quite negligible quantity of foreign material, there was considered to be little danger of toxic effects.

Design and dimensions of measuring prosthesis

Prosthetic head

The size of the prosthetic head should accurately fit the acetabulum. In ordinary arthroplastic operations in the hip-joint, different sizes of prostheses are available. As a limited number of measuring prostheses were to be manufactured the head of the prosthesis was made $1^7/_8''$ or 47.2 mm in diameter which, according to our experience, is the size most commonly used. The prosthesis has been made somewhat larger than half a sphere. To obtain enough length in the neck part — the available space is limited for anatomical reasons — the head was made hollow. In addition, this enabled the centre of the head to be placed 10 mm into the neck, and strain gauges could be applied in a section through the centre of the prosthetic head.

Neck of prosthesis.

Recording of strain takes place in the neck part by means of applied strain gauges. To obtain satisfactory measuring accuracy the neck of the prosthesis must be rather long. Available space from an anatomical viewpoint appears to be about 40 mm. Rigidity against radial deformations in the front part of the neck is obtained by a sturdy flange. This flange is threaded and has fittings for the prosthetic head. As the neck is hollow and the strain gauges placed on the inside, it must be possible to pull out the wires. The flange was made with a hole for the nipple and lead wires.

To fit into the stem of the prosthesis, the neck part was provided with a cone-shaped shank with internal thread. By such means it was possible to obtain a joint with zero slackness and high load bearing capacity and with a minimun neck length. The base of the neck tube was closed by a metal plate to prevent radial deformations.

For technical reasons the included angle between the axis of the neck part and the stem was made 120°. This is 6° less than the average cervico-diaphyseal angle (Backman 1957). In the hip the angle between the prosthetic neck and stem will represent the angle between the neck and the longitudinal axis of the proximal femur. The true cervico-diaphyseal angle, however, is formed between the neck and the ideal axis of the femur and this angle is somewhat greater. The difference is partially dependent on the magnitude of antetorsion the prosthesis will have in the hip. The internal diameter of the neck was determined on the basis of the installation of the strain gauges. 16 mm is a suitable diameter, both as regards the glueing of the gauges and the anatomical conditions.

The section thickness of the prosthesis depends on the strength of the material and the force expected to act on the head. Providing force P acts through the centre of the prosthetic head, the moments caused by the vertical and horizontal components P_y and P_z are obtained in a section through the prosthetic neck from expressions:

$$P_y = \frac{M_{2z}}{L} \tag{16}$$

and

$$P_z = \frac{M_{2y}}{L} \tag{17}$$

where L is the distance between the mid-point of the head and the measuring section in the prosthesis, designated section 2.

For the purpose of dimensioning, these expressions can be given the formula:

$$M_{2z} = L \cdot P \sin \alpha^{*} \tag{18}$$

and in bending, according to Hooke's law:

$$LP \sin \alpha = E \cdot W_{b2} \cdot \varepsilon_{b2} \cdot \frac{D_2}{d} \tag{19}$$

where E is the modulus of elasticity and ε_{b2} = the bending strain of the inner surface in the sectional cross-section.

Further:

$$W_{b2} = \frac{\pi \cdot (D_2^4 - d^4)}{32 \; D_2} \tag{20}$$

where D_2 = major sectional diameter.

and d = minor sectional diameter.

If a strain of $\varepsilon_{b2} = 1$ ‰ occurs for a force P of 400 kp acting on the head and $\alpha = 60°$ and $\gamma = 0$, a deflexion of the galvanometers of 12 cm takes place. Thus, $\varepsilon_{b2} = 1$ ‰ has been considered suitable and in that case $D_2 = 18.65$ mm and thickness in section 2 $t_2 = 1.33$ mm. The thickness of the prosthetic neck was settled to 1.4 mm in section 2.

If no friction occurs in the joint, force P acts through the centre of the head and the moments in section 1; M_{1z} and $M_{1y} = 0$. To determine the thickness in this section, calculations must be based on the smallest

* Of course, this formula applies only if $\gamma = 0$.

section thickness, which is obtained next to the fillet 8 mm in front of section 1. Analogously with the expression for section 2 for a section next to the fillet is obtained:

$$-8\ P \sin \alpha = E \cdot W_{bf} \cdot \varepsilon_{bf} \cdot \frac{D_f}{d} \tag{21}$$

where ε_{bf} is equivalent to the bending strain of the inner surface in the cut and

$$W_{bf} = \frac{\pi\,(D_f^4 - d^4)}{32\ D_f} \tag{22}$$

If D_f = external diameter of the cut and if $\varepsilon_{bf} = 1\ ^0/_{00}$, P = 400 kp and $\alpha = 60^\circ$, D_f must be 17.2 mm and $t_f = 0.6$ mm.

The thickness of the prosthetic neck next to the fillet was fixed at 0.7 mm. With 0.7 mm at the fillet and 1.4 mm at section 2, in section 1 the thickness $t_1 = 0.9$ mm is obtained.

The gauges measuring the axial force P_x was placed between the two sections at a distance of 8 mm from section 1. In this section the thickness $t_3 = 1.1$ mm and A_3 (cross-section area) = 59.2 mm².

Strain due to the axial force is obtained from expression:

$$\varepsilon_{p3} = \frac{P \cos \alpha}{A_3 \cdot E} \tag{23}$$

If P = 400 kp and $\alpha = 60^\circ$, $\varepsilon_{p3} = 0.165\ ^0/_{00}$.

With the dimensions set forth above the maximum permissible loads have been calculated. With regard to static loads, the greatest stress concentration will occur at the lateral fillet. The nominal stress in this cut is obtained according to:

$$\sigma\ nom_f = \pm \frac{P \sin \alpha\,(L+7.5)}{W_{bf}} - \frac{P \cos \alpha}{A_f} \tag{24}$$

where A_f = cross-section area and

$$W_{bf} = \frac{\pi\,(D_f^4 - d^4)}{32 \cdot D_f} \tag{25}$$

where D_f = major diameter of cut and d = minor diameter of the cut. If P = 400 kp and α 60° $\sigma\ nom_f = -28{,}5$ kp/mm².
is obtained.

For the fillet in question a form factor of 1.6 can be given, which means that $\sigma\ max_f = -45.5$ kp/mm².

With the given strength data, local plastic creeping will occur in the fillet for static loads exceeding 570 kp and fractures for static loads exceeding 920 kp.

The strength-decreasing effect of the fillet will, however, be eliminated by the stress adjustment occurring due to creeping of the material, so that fractures will not occur unless the static loads are considerably greater than the maximal loads indicated above.
The material has a minimal fatigue strength of ± 25 kp/mm^2 amplitude for a lifetime of 10^7 cycles of loading and is increased by local stress adjustment.
Since it is mainly a question of pulsating loads, an upper limit of about 40 kp/mm^2 is permissible. If a deduction of 20 % is made for the strength-decreasing factors, a fatigue strength of up to 32 kp/mm^2 is permitted. As the material has no notch brittleness, the margin of safety should be sufficient if a notch fatigue factor of 1.2 is considered. Fractures due to fatigue will not occur until loads exceed 390 kp.

Stem of prosthesis.
The only difference between the stem of the measuring prosthesis and that of a conventional Moore prosthesis is its somewhat larger dimensions.

The prosthesis.
The measuring prosthesis has three parts: the head, the neck and the marrow space anchorage (fig. 9).
In addition, there were internal gauges, from which leads passed out through a nipple sunk into the flange of the neck. At the distal end the leads, 22 cm long, were joined to a multi-pole female contact built into a hermetically-sealed metal cylinder. The passage of the leads through the nipple was sealed with resin. The leads, insulated separately, also passed through a Teflon collar threaded on the nipple. In order to give the cable formed by the wires better rigidity, strips of Teflon were placed between the various wires. In this way there was less risk of the wire-insulation cracking. The female contact contained toxic material and was placed in

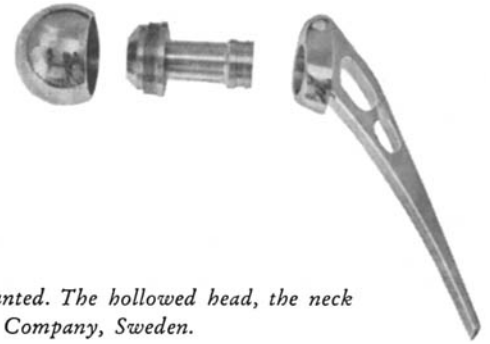

Fig. 9. *The measuring prosthesis unmounted. The hollowed head, the neck part and the stem can be seen. Bofors Company, Sweden.*

a hermetically-sealed cylinder. This cylinder was made of the same material as the prosthesis. Also the nipple of the cylinder was sealed with resin.

The wires going out from the prosthesis were removed when the measurements had been completed. It is desirable to remove the wires as close to the passage in the prosthesis as possible. The nipple and the threaded-on Teflon collar were encased in a steel sleeve glued to the prosthesis. The purpose of the steel sleeve was to prevent cracking of the wires as they pass out from the nipple, to improve the sealing and to serve as a sharp edge to cut the wires when they were to be removed. By pulling the wires against the sharp edge of the sleeve it was hoped that they would be severed. In practice this did not occur and an incision had to be made to remove the wires.

The threaded units of the prosthesis were fixed by means of resin. For gluing the strain gauges, wires, the lateral threaded unit and nipple, a resin of the Araldite XV type, which was heat-cured at 150°, was used. An Araldite D type resin was used as adhesive for the strain gauges for component M_x. The mounting of the prosthetic head, which was the last phase of the installation, was carried out with Araldite D with Hy 956 activator. As this is a room temperature curing adhesive, it was unnecessary to heat the measuring prosthesis during the hardening process. With a high-temperature curing resin it is very difficult to obtain an absolutely tight unit because the air in the measuring prosthesis expands with heat and forms canals in the glue joint.

The main parts of the prosthesis were tested with ultra-sound for cracks. To obtain the overall surface finish the head and the stem were electropolished. Fig. 10 shows the measuring prosthesis mounted in its finished condition. Fig. 11 is an outline drawing of the prosthesis giving dimensions.

Measuring receptors and measuring principles

In the first prosthesis Hottinger strain gauges of type 3/120 FB1 have been used for recording the components M_{1z}, M_{2z}, M_{1y}, M_{2y} and P_x. The gauges have an active length of 3 mm and a resistance R of 120 ohm. The gauging factor g is 1.99 ± 1 %.

The second prosthesis has been provided with Baldwin strain gauges of type AB-11 with a resistance of 120 ohm and a gauging factor of 1.90 ± 1 %. The active length of the gauges is 3 mm. For component M_{1x} another strain gauge, of Budd manufacture type C6-121-R_{3a}, has been used. Its resistance is 120 ohm, the gauging factor 1.99 ± 1 %, active length 3 mm. Theories concerning measuring with strain gauges can be found in litera-

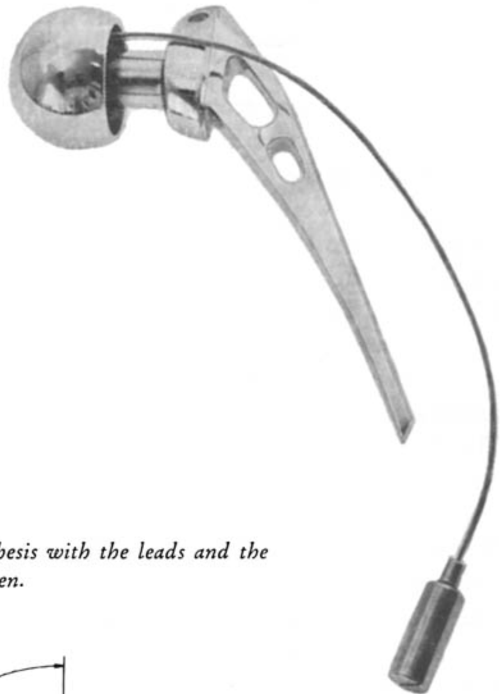

Fig 10. *The mounted measuring prosthesis with the leads and the contact house. Bofors Company, Sweden.*

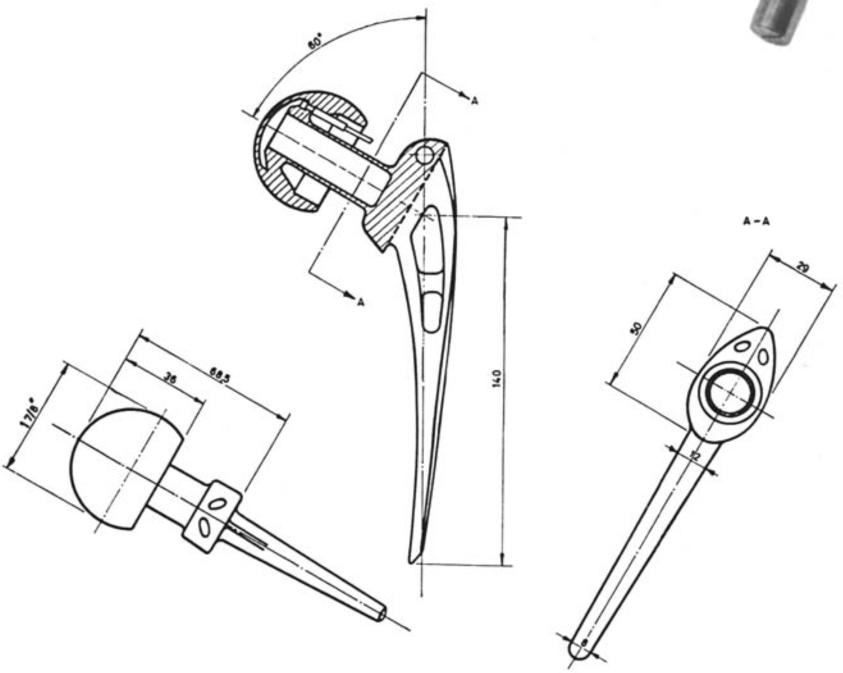

Fig. 11. *An outline drawing of the measuring prosthesis. Figures in mm when not indicated otherwise.*

ture on this subject (e. g. *Mechanical Measurements,* Beckwith and Buck 1961, *The Strain Gage Primer,* Perry and Lissner 1962 and *Instrument Transducers,* Neubert 1963). The measurements are based on changes in resistance of the gauges, which are connected in a Wheatstone bridge. If one resistance is altered this will result in imbalance in the bridge and, in a suitable way, this can be recorded as a change of voltage of the bridge. For a Wheatstone bridge, an imbalance must arise in proportion to the algebraic difference of resistance changes in any two adjacent arms, and in proportion to the algebraic sum of resistance changes in any two opposite arms. With strain gauges — strains cannot only be measured, but through suitable coupling the recorded strains can be eliminated or enlarged. Suitable placing of the strain gauges in the Wheatstone bridge

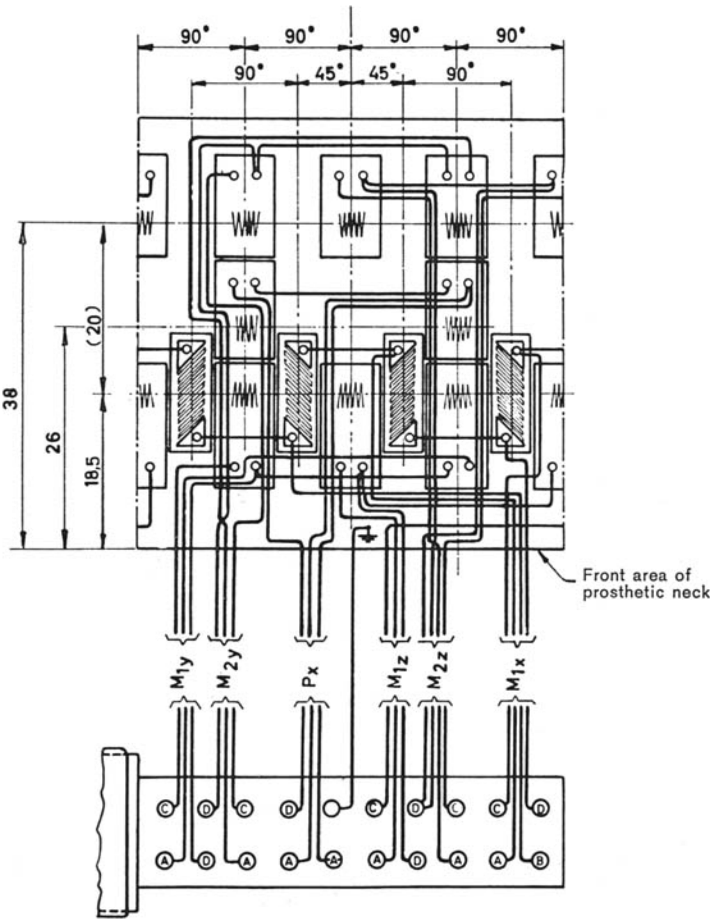

Fig. 12. *The coupling scheme of the prosthesis of case 2.*

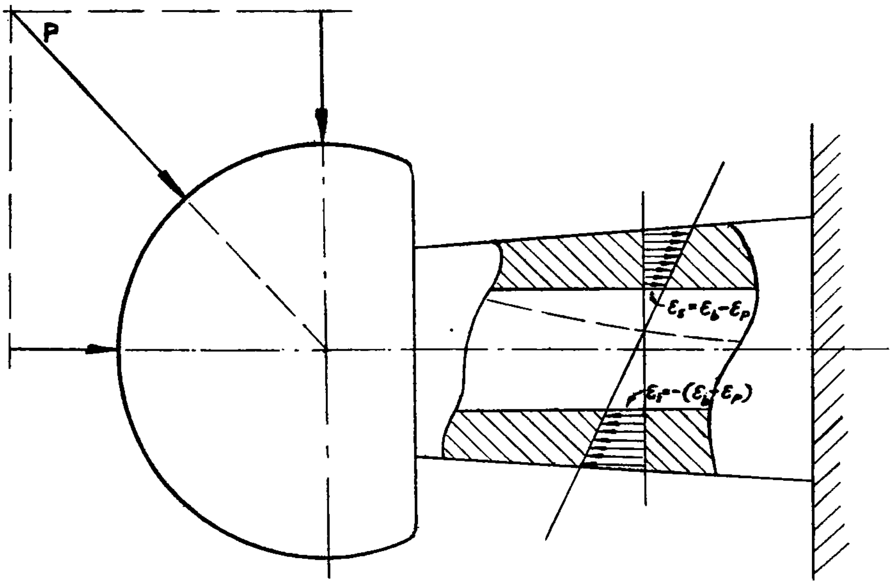

Fig. 13. *The stress distribution in a neck section when the head is subjected to a force P. Both tensile and compressive strains will occur. (ε=strain, s=superior, i=inferior, b=bending and p=pressure).*

allows forces to be separated from moments. According to the same theories, the torque around the longitudinal axis of a beam can be determined.

In accordance with these principles the strain gauges have been applied in the prosthetic neck. Due to the space available, the prosthesis has been provided with four half-bridges for determination of the moment and two strain gauges placed in series for determining the axial force. Fig. 12 shows the coupling scheme for prosthesis 2. The connection to mid-point of the M_{2z}-bridge has been doubled. The same coupling scheme applies to prosthesis 1 with the exception of gauges for M_{1x}. In an outer adapting unit the strain gauges are supplemented so that five complete bridges are obtained. The adapting unit is provided with a waist-belt. It is to be strapped around the waist so as to cause as little discomfort as possible. Fig. 13 shows the stress distribution in a section where the prosthetic head is subjected to force P. Both tensile and compressive strains will occur in the prosthetic neck. The strain measured at one point accordingly includes both tensile and compressive components. These are separated by subtracting or adding the strains in two diametrical points.

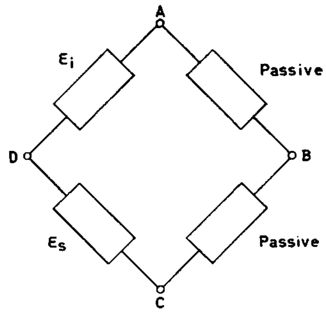

Fig. 14. *Differential coupling to obtain strains from bending.*

The strain caused by the bending moment is obtained by means of differential coupling as per fig. 14. Thus, the expression is obtained:

$\varepsilon_s - \varepsilon_i = 2\varepsilon_b$ (s = superior, i = inferior, ε_b = strain from bending) (26)

where ε_s has a positive sign and ε_i a negative sign.
The strain caused by the axial force is obtained with coupling per fig. 15. Thus, the expression is obtained:

$$\frac{\varepsilon_s + \varepsilon_i}{2} = -\varepsilon_p \quad (\varepsilon_p = \text{strain from pressure}) \qquad (27)$$

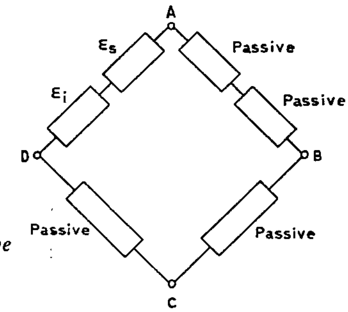

Fig. 15. *Coupling to obtain strains from compressive forces.*

The strain gauges for determining M_{1x} have been placed in section 1. These gauges are connected in a complete Wheatstone bridge as shown in fig. 16, and thus the expression is:

$$\Sigma\varepsilon = 4 \cdot \varepsilon_{tr} \quad (\text{tr} = \text{torsional strain}) \qquad (28)$$

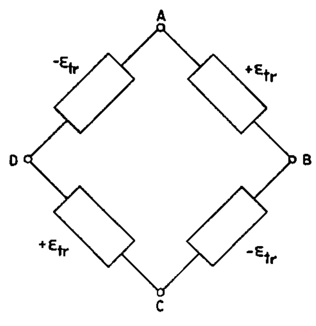

Fig. 16. *Coupling to obtain torsional strains.*

Temperature compensation is unnecessary for the gauges as the measuring occurs under conditions which are stable as regards temperature. Difficulties can arise, however, with the temperature drift for the lead wires. For all moments this is compensated by the position of the gauges and the leads in the bridge. For P_x the so-called Ruge-coupling has been used (fig. 17), which eliminates the temperature drift of the lead wires. The temperature sensitivity of the applied gauges is about 0.03 % per degree C.

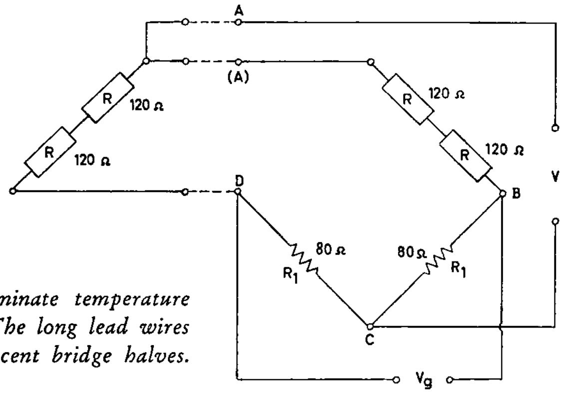

Fig. 17. *Ruge-coupling to eliminate temperature drift of the lead wires of P_x. The long lead wires must be connected to two adjacent bridge halves.*

The sensitivity of the measuring prosthesis

The amplitudes corresponding to the load shown on the oscillograph film are determined as follows:

Applying for the moment sections:

$$a_m = \frac{V}{2} \cdot g \cdot \varepsilon_b \cdot c \tag{29}$$

where V=bridge voltage, g=gauge factor, ε_b=bending strain, c=sensitivity of galvanometer. If a maximal bridge voltage of 6 volts is permissible, at $\varepsilon_b = 1$ ‰ an amplitude of $a_m = 11.3$ cm is obtained.

It is still assumed that no signals are received from the gauges measuring moments in section 1, which applies if the load is transmitted without friction.

For the section in which the axial load P_x is measured, the following applies:

$$a_{p3} = \frac{V}{4} \cdot \frac{R_{BD}}{R_{AC}} \cdot g \cdot \varepsilon_{p3} \cdot c \tag{30}$$

where R_{BD}=resistance between the bridge corners BD, and R_{AC}=resistance between the bridge corners AC . ε_{p3}=the axial strain.

As shown earlier, $\varepsilon_{p3}=0.165$ ‰ if P=400 kp and $\alpha=60°$.

If a maximal bridge voltage of 12 volts is permissible, a galvanometer deflection $a_{p3} = 1.4$ cm is obtained.
In regard to the torsion:

$$a_{tr} = V \cdot g \cdot \varepsilon_{tr} \cdot c \qquad (31)$$

where

$$\varepsilon_{tr} = \frac{M_{1x}}{E} \cdot \frac{(1+\mu)}{W_{tr}} \cdot \frac{d}{D_1} \qquad (32)$$

where

$$\mu = 0.3 \text{ (Poisson's ratio)}$$

and

$$W_{tr} = \frac{\pi (D^4_1 - d^4)}{16 \cdot D_1} \qquad (33)$$

With 0.9 mm section thickness (t_1) and a torque of 500 kpmm (max.-moment obtained for components M_{1y} and M_{1z} when measuring with prosthesis No. 1), a galvanometer deflexion $a_{tr} = 1.7$ cm is obtained if a max. bridge voltage of 6 volts is permissible.

Calculation of calibration constants

For calibration, a known resistance R_K is connected across the bridge arm BC. The unbalance thereby obtained is related to the unbalance obtained by loading. The following connection applies if R_K is very much greater than R.

$$\frac{\Delta R}{R} = \frac{R}{R_K} \qquad (34)$$

If the resistance change, which is obtained when R_K is put in circuit, equals total resistance change in bridge when loaded, the expression is obtained:

$$\frac{\Delta R}{R} = \frac{R}{R_K} = g \cdot \Sigma\varepsilon \qquad (35)$$

thus

$$\Sigma\varepsilon = \frac{R}{g \cdot R_K} \qquad (36)$$

The following applies for the moment sections:

$$\Sigma\varepsilon = 2\varepsilon_b \qquad (37)$$

thus

$$\varepsilon_b = \frac{R}{2 \cdot g \cdot R_K} \qquad (38)$$

For moment section $2\varepsilon_b$ in the foregoing expression (38) can be inserted in the following:

$$LP \sin \alpha = E \cdot W_{b2} \cdot \varepsilon_{b2} \cdot \frac{D_2}{d} \qquad (39)$$

and as a result

$$M_{2K} = E \cdot W_{b2} \cdot \frac{R}{2 \cdot g \cdot R_K} \cdot \frac{D_2}{d} \qquad (40)$$

The moments corresponding to a number of standard values for R_K are listed in table 2.

R_K	Prosthesis 1 M_{2K}	Prosthesis 2 M_{2K}
80 kΩ	2810 kpmm	2944 kpmm
160 kΩ	1405 kpmm	1472 kpmm
320 kΩ	703 kpmm	736 kpmm

Table 2. *The moments corresponding to known resistance R_k. Section 2.*

Analogous with the expression for moment section 2 the following is obtained for section 1:

$$M_{1K} = E \cdot W_{b1} \cdot \frac{R}{2 \cdot g \cdot R_K} \cdot \frac{D_1}{d} \qquad (41)$$

giving the results presented in Table 3.

R_K	Prosthesis 1 M_{1K}	Prosthesis 2 M_{1K}
160 kΩ	828 kpmm	864 kpmm
320 kΩ	414 kpmm	432 kpmm
640 kΩ	207 kpmm	216 kpmm

Table 3. *The moments corresponding to known resistance R_k. Section 1.*

For section 3 in which the axial force P_x is measured:

$$\Sigma\varepsilon = \varepsilon_{p3} \qquad (42)$$

and analogous with expression for moment sections is obtained:

$$\varepsilon_{p3} = \frac{R_1}{g \cdot R_K} \qquad (43)$$

If formula for ε_{p3} in this expression is inserted in the expression:

$$\varepsilon_{p3} = \frac{P \cos \alpha}{A_3 \cdot E}$$

$$\text{and as } P \cos \alpha = P_x \qquad (45)$$

this expression is obtained:

$$P_x = E \cdot A_3 \cdot \frac{R_1}{g \cdot R_K} \qquad (46)$$

From this equation the values for P_xK presented in Table 4, corresponding to R_K are obtained:

R_K	Prosthesis 1 P_xK	Prosthesis 2 P_xK
160 kΩ	304 kp	318 kp
320 kΩ	152 kp	159 kp
640 kΩ	76 kp	80 kp

Table 4. *Values of P_x corresponding to known resistances R_k.*

The following applies to the torsion:

$$\Sigma\varepsilon = 4\,\varepsilon_{tr} \tag{47}$$

and if

$$\Sigma\varepsilon = \frac{R}{g \cdot R_K} \tag{48}$$

the expression is obtained:

$$\varepsilon_{tr} = \frac{R}{4 \cdot g \cdot R_K} \tag{49}$$

The expression for ε_{tr} is inserted in equation 32, thus resulting in

$$M_{1x}K = \frac{E \cdot W_{tr}}{1+\mu} \cdot \frac{R}{4 \cdot g \cdot R_K} \cdot \frac{D_1}{d} \tag{50}$$

The moments corresponding to a number of standard values for R_K have been determined and values obtained can be seen in Table 5.

R_K	$M_{1x}K$
160 kΩ	635 kpmm
320 kΩ	318 kpmm
640 kΩ	159 kpmm

Table 5. *The torsional moment corresponding to known resistances R_k.*

The theoretical calibration constants are presented in tables 6 and 7 together with measured calibration constants.

Calibration of the measuring prosthesis

The calibration method was identical for both prostheses, except as regards the torsion moment, as the gauges for this have only been applied in prosthesis 2.

The socket of the contact was removed and the measuring prosthesis was connected to the adapting unit (fig. 18). The outgoing wires of the adapting unit were connected respectively to a 5 and 6 channel balancing and calibration unit which, via a switch, was connected to a signal amplifier and digital voltmeter.

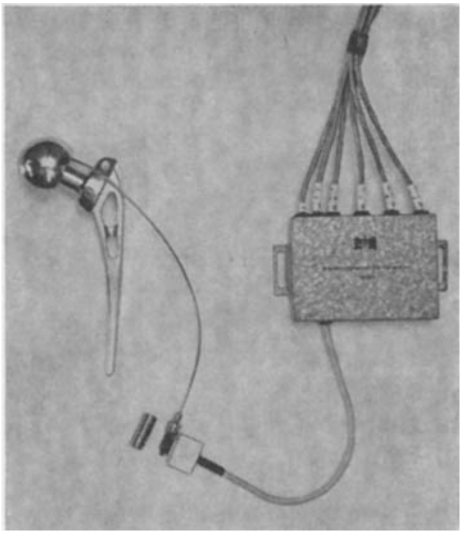

Fig. 18. *The measuring prosthesis connected to the adapting unit. Bofors Company Sweden.*

Calibration without the prosthetic head. (With adaptor)

In order to obtain as complete calibration data as possible, prosthesis 1 and 2 were calibrated before the head was mounted. For this purpose a metal cylindrical socket provided with V-shaped grooves with ball seatings was manufactured. This adaptor was threaded to the neck in place of the head and the prosthesis could be loaded at defined points. So that the force acting on the prosthesis would be accurately directed, suspended weights were used. The load was transmitted to the prosthesis via a steel ball placed in the aforementioned ball seatings. The prosthesis was secured in a holder of Resitex (fig. 19).

The angles used for describing the directions of the forces are defined on page 29.

The calibration of the M_z sections was performed with $\alpha=90°$, $\gamma=0°$ and $\beta=90°$, the calibration of the M_y sections with $\gamma=90°$ and $\beta=90°$. The torsion moment was calibrated via a lever. The gauges measuring the force P_x were calibrated with $\alpha=0°$ and $\beta=0°$.

Calibration with prosthetic head.

Same procedure for both prostheses, except that prosthesis 2 was calibrated with and without the head glued. After mounting and fixing the head, calibration was again carried out. It was thus possible to get an idea as to wether the midpoint of the moment section 1 deviated from the centre of the prosthetic head, and whether the calibration constants were changed when the head was secured.

Component	M_{1z}	M_{2z}	M_{1y}	M_{2y}	P_x
R_k (kΩ)	320	160	320	160	320
		kpmm			kp
Prosthesis 1					
Theoretical	414	1405	414	1405	152
With socket	470	1470	460	1460	171
With Ball	—	1450	—	1455	170
Definitive	470	1460	460	1460	170
Lp_y = 20.8 mm					
Lp_z = 20.5 mm					

Table 6. *Results of calibration prosthesis 1. Constants and distances between section 1 and 2 are given.* L_{py}*=the lever arm of the component* P_y. L_{pz}*=the lever arm of the component* P_z.

Component	M_{1z}	M_{2z}	M_{1x}	M_{1y}	M_{2y}	P_x
R_k (kΩ)	320	160	320	320	160	320
		kpmm				kp
Prosthesis 2						
Theoretical	432	1472	318	432	1472	159
With socket	470	1520	376	460	1525	174
With Ball unglued and glued	—	1520	376	—	1520	174
Definitive	470	1520	376	460	1520	174
Lp_y = 20.9 mm						
Lp_z = 20.8 mm						

Table 7. *Results of calibration prosthesis 2. Constants and distances between section 1 and 2 are given.* L_{py}*=the lever arm of the component* P_y. L_{pz}*=the lever arm of the component* P_z.

The calibration constants and the distance between the two moment sections obtained from calibration tests are indicated in tables 6 and 7. Satisfactory agreement was obtained for the calibration constants measured in various ways. Disagreement between the theoretical and measured values was largely due to the rather high ohm values of the lead wires which had been disregarded in the calculations.

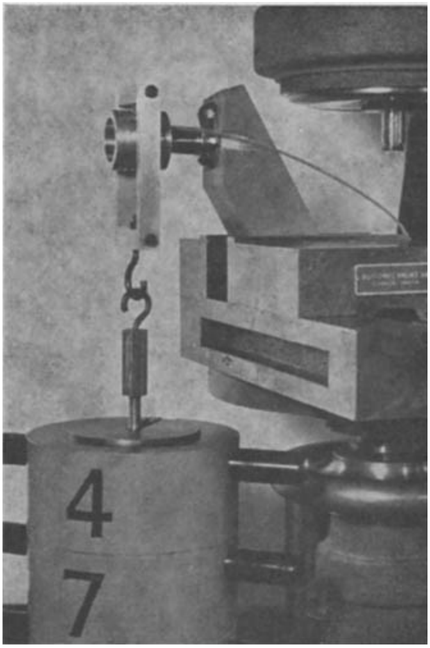

Fig. 19. *Calibration of the prosthesis via a cylindrical adaptor. Bofors Company Sweden.*

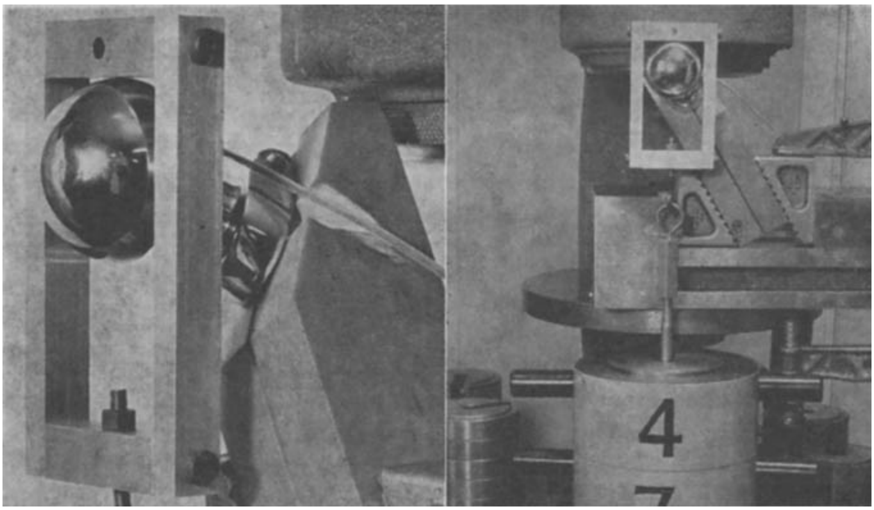

Fig. 20. *Control loading of the prosthesis. Bofors Company Sweden.*

Control loading

To test the calibration constants obtained, the measuring prostheses were loaded under two different conditions with 40 and 80 kp, respectively. Forces and angles were calculated from the recordings thus obtained. These were compared with the actual loads and angles. By this means an idea of the measuring accuracy was obtained. The components P_x, P_y and P_z were also compared.

The following trigonometric connections exist:

$$P_y = P \sin\beta \cos\gamma \tag{51}$$

$$P_z = P \sin\beta \sin\gamma \tag{52}$$

$$P_x = P \cos\beta \tag{53}$$

When positioning the prosthesis at given angles, angles α and γ can be checked.

If β is expressed in γ

$$P_y = P \cos\gamma \sqrt{\frac{1}{1 + \left(\frac{\cos\gamma}{\tan\alpha}\right)^2}} \tag{54}$$

$$P_z = P \sin\gamma \sqrt{\frac{1}{1 + \left(\frac{\cos\gamma}{\tan\alpha}\right)^2}} \tag{55}$$

$$P_x = P \sqrt{\frac{1}{1 + \left(\frac{\tan\alpha}{\cos\gamma}\right)^2}} \tag{56}$$

Fig. 20 shows how the load was transmitted to the prosthetic head. If the plane transmitting the force is correctly adjusted in the horizontal plane the load is bound to act via the centre of the head.

Control loadings with prosthesis 1.

The prosthesis was secured in such position that $\alpha = 60^\circ$, $\gamma = 33^\circ$, and was subjected to a load of 40 kp. The position was then changed to $\alpha = 60^\circ$ and $\gamma = 35^\circ$ and a load of 80 kp was applied. Table 8 shows the recordings obtained.

Measuring section			M_{1z}	M_{2z}	M_{1y}	M_{2y}	P	
Bridge voltage			6	6	6	6	12	R_K
P	α	γ		vg x 500				kΩ
			mV	mV	mV	mV	mV	
			584	587	590	588	562	160
			292	294	295	294	281	320
40	60°	33°	—10	245	5	161	28	
80	60°	35°	—18	488	10	334	56	

Table 8. *Recordings of control loading, prosthesis 1.*

Table 9 shows a comparison between the measured and the actual values of forces and angles. Corrections have been done for considering mutual interaction.

	P_y kp	P_z kp	P_x kp	P kp	α	γ	β
Applied load and angles	30.2	19.7	17.4	40.0	60°	33°	64.1°
Measured uncorrected	29.1	19.4	17.0	38.9	59.8°	33.6°	64.1°
Measured corrected	29.5	20.5	17.0	39.8	60.2°	34.8°	64.7°
Applied load and angles	59.2	41.5	34.2	80.0	60°	35°	64.7°
Measured uncorrected	58.0	40.2	33.9	78.2	59.7°	34.7°	64.3°
Measured corrected	58.5	41.4	33.9	79.3	59.9°	35.3°	64.7°

Table 9. *Applied and measured forces and angles of prosthesis 1.*

The percentual error for measured forces and angles is shown in table 10:

		Error in %			
		P	α°	γ°	β°
P=40 kp	uncorrected	—3.0	—0.4	2.0	0.0
α=60° γ=33°	corrected	—0.5	0.4	5.5	1.0
P=80 kp	uncorrected	—1.0	—0.5	—0.9	—0.7
α=60° γ=35°	corrected	—0.9	—0.2	0.9	0.0

Table 10. *The procentual error of measured forces and angles, prosthesis 1.*

Control loadings with prosthesis 2

The control loadings for prosthesis 2 were carried out similarly to those for prosthesis 1, and with the same equipment. The prosthesis was secured in such position that $\alpha=60°$ and $\gamma=35°$. In table 11 the measured and the actual values of forces and angles are presented.

	P_y kp	P_z kp	P_x kp	P kp	α	γ	β
Applied load and angles	30.2	19.7	17.4	40	60°	33°	64.1°
Measured uncorrected	30.8	19.8	17.4	40.5	60.5°	32.7°	64.5°
Measured corrected	30.3	19.8	17.4	40.2	60.1°	33.1°	64.3°
Applied load and angles	52.8	34.4	30.4	70	60°	33°	64.1°
Measured uncorrected	54.0	34.4	30.5	71	60.5°	32.5°	64.5°
Measured corrected	53.2	34.6	30.5	70.5	60.2°	33°	64.4°

Table 11. *Applied and measured forces and angles of prosthesis 2.*

Table 12 shows the error in % of measured forces and angles.

		P	$\alpha°$	$\gamma°$	$\beta°$
P=40 kp	uncorrected	1.3	0.8	—1	0.6
$\alpha=60°$	corrected	0.5	0.2	0.3	0.3
$\gamma=33°$					
P=70 kp	uncorrected	1.4	0.8	—1.5	0.6
$\alpha=60°$	corrected	0.7	0.3	0	0.5
$\gamma=33°$					

Table 12.

The moment around the x-axis was tested with socket and with the prosthetic head and with different lengths of lever arms. In all tests 50 kpmm corresponded to 20 mV, 100 kpmm to 40 mV, 200 kpmm to 80 mV etc.

Recording of dynamic strains

Strain gauges can be used for determining dynamic strains. The gauge itself will respond to strains of frequencies of more than 50,000 cps.

If the prosthesis has a natural frequency close to the frequencies which are generated, this might influence the recordings. The natural frequency and spring constant of the prosthesis were therefore determined. The natural frequency proved to be f=2 500 cps.

According to expression:

$$c = m\,(2\,\pi\,f)^2 \qquad (57)$$

the constant c=5 000 kp/mm is obtained.

Hence, the prosthesis has a stiffness considerably greater than the stiffness of the parts which it replaces and is surrounded by. The prosthesis is an integral part of mass-spring unit and therefore cannot affect the dynamic quality of the unit so that too small forces are measured. Furthermore, it will correctly record the occurring forces irrespective of the force-frequency.

The strains are recorded by means of a Honeywell visicorder model 906 T. The galvanometers used respond to frequencies of up to 60 cps with an accuracy of ± 5 %. Low frequencies will be recorded with greater accuracy, and when measuring forces on the hip-joint, frequencies higher than 6 cps will probably not occur.

Friction in the hip-joint

Under load the surfaces of the joint will press against each other with a certain normal stress. Motion between two surfaces will cause tangential friction stresses.

The size of the stresses depends on the magnitude of the applied load, and the quality of the contact surfaces. The direction of the tangential stresses depends on the motion. The ratio between tangential and normal stress is called the coefficient of friction μ and is dependent on the quality of the contact surfaces and on the lubrication, and sometimes on the magnitude of normal stress. In simple cases where the surfaces are flat and μ is a constant, the ratio between the resultant friction force and the resultant normal force will have the value μ independent of the size of the contact area, the coefficient of friction may be determined when the frictional force is limiting from the expression:

$$\mu = \frac{f}{n} \qquad (58)$$

where f is the max. force of friction and n the force of reaction perpendicular to the contact surface. This is approximately valid for a curved surface like a sphere too, if the normal stresses are distributed over a small area only.

In an animal joint the coefficient of friction has a low value. On autopsy specimens Jones (1936) and Charnley (1959, 1960) determined the coefficient of friction in a joint, and obtained values between 0.013 — 0.02.

Barnett and Cobold (1962) tried to determine the coefficient of friction under living conditions. Tests were made in a finger joint and the authors obtained a min. value of 0.0075. The authors also discovered that the coefficient of friction will decrease with increasing load — a property also seen in certain types of plastic material. Under dynamic conditions it is an advantage if the coefficient of friction diminishes with increasing load, whereas under static load the opposite condition would be advantageous. Hirsch (1944) demonstrated that under static load the shape of the joint-surfaces are affected, which might increase the coefficient of friction.

In cases where one part of a joint is replaced by a construction of metal, a different coefficient of friction may be expected.

In the preceding chapter it has been assumed that force P acting on the hip-joint is directed through the centre of the prosthetic head. If friction

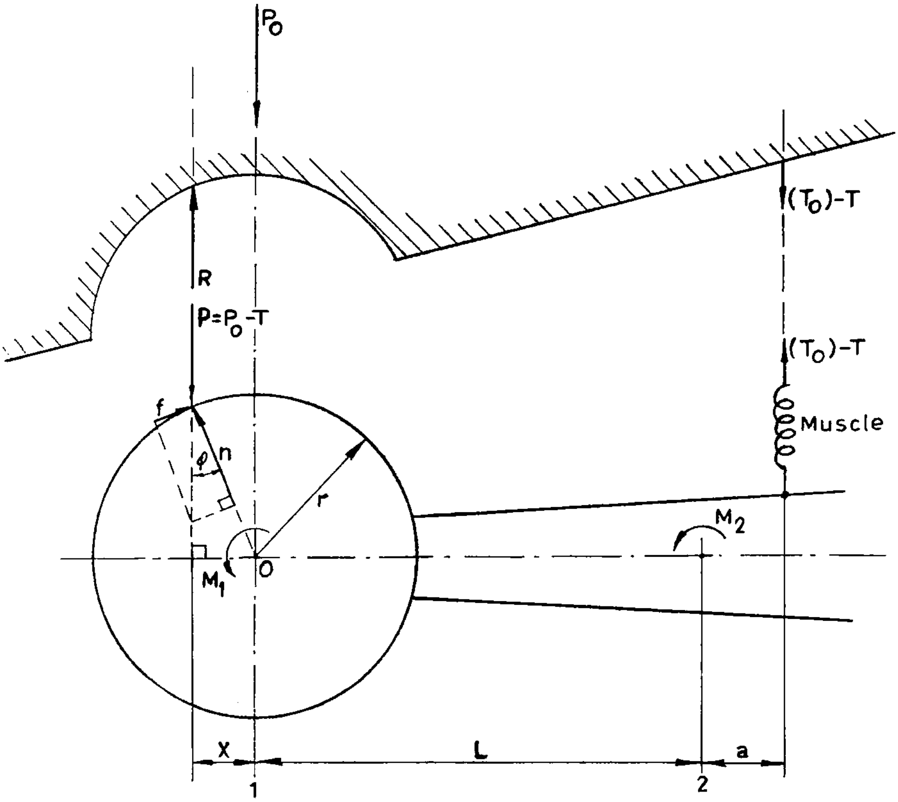

Fig. 21. *Due to friction a force P_0 is displaced from the origin if the muscular force is changed by an amount T. The force of reaction R will result from the normal force n and the friction force.*

occurs, the combination of a normal force and frictional force can be described as if the resultant is displaced and will maintain equilibrium until the frictional force is limiting. In the case of static loads such a frictional force may arise if the muscular force is increased or decreased (fig. 21).

The coefficient of friction μ may be determined with the measuring prosthesis when the frictional force is limiting. For the case of friction between two curved surfaces approximately the same laws apply as for an inclined plane if the contact area is not too great. For the moment around the origin we have:

$$M_1 = P \cdot x \tag{59}$$

Fig. 21 shows that:

$$x = r \cdot \sin \varphi \tag{60}$$

r = the radius of the prosthetic head.

For small angles the sine is approximately equal to the tangent and to the angle expressed in radians. As we are concerned with small angles, we can put $\sin \varphi = \tan \varphi$. Moreover, since $\tan \varphi = \frac{f}{n}$ where f is the frictional force and n the normal force acting perpendicular to f, $\tan \varphi =$ the coefficient of friction μ.

Thus, $M_1 = P \,.\, r \,.\, \mu$

but

$$M_1 = \sqrt{M^2_{1z} + M^2_{1y} + M^2_{1x}} \tag{61}$$

and the expression is obtained:

$$\mu = \frac{\sqrt{M^2_{1z} + M^2_{1y} + M^2_{1x}}}{r \,.\, P} \tag{62}$$

This formula is valid provided that the stresses are distributed over a rather small area and the radius of friction is equivalent to the radius of the prosthetic head, irrespective of the direction of the force P and distribution of the moment of friction between the axes of x, y and z. However the friction radius is to some extent dependent on these factors and on the position of the prosthetic head in the acetabulum, so the equation may not be exactly correct. Since the friction radius can assume values that are either greater or less than the radius of the prosthetic head, the latter is an acceptable mean value of the relvant radii of friction. Hence, the above expression for the coefficient of friction is probably acceptable for obtaining approximate values.

The torsional moment M_{1x} need not be known to determine the magnitude and direction of P. Therefore, even if M_{1x} is not measured it should be possible to obtain an approximate value for the coefficient of friction by determining μ from the resultant of M_{1z} and M_{1y}. Since the coefficient of friction is to be determined when this resultant is a maximum, it may be assumed that M_{1x} will be small and therefore not affect the results. On the other hand, if M_{1x} is of the same magnitude as

$$\frac{M_{1z} + M_{1y}}{2} \tag{63}$$

an error is incurred which, however, will not exceed $\sqrt{\frac{3}{2}} - 1 = 20\,\%$.

In prosthesis 1 no gauges were applied measuring M_{1x}, but in prosthesis 2 the moment M_{1x} around the longitudinal axis of the neck part was recorded.

If the contact area is too great, this method introduces an error, which gives too high values of the coefficient of friction.

Surgical viewpoints

Attachment

The aforedescribed measuring prosthesis shall be fixed to the femur in the same way as a conventional prosthesis. This implies that the neck of the femur must have a suitable angle and that the marrow space anchorage of the prosthesis must be forced down via the neck stump into the femoral marrow space. It is important that the prosthetic head and neck do not receive any hard blows when secured into the femoral marrow space. In that event, the measuring receptors could be damaged. To facilitate insertion of the prosthesis, a reamer has been manufactured. Having the same dimensions as the prosthetic stem, and when inserted into the marrow space it will form a suitable bed for the prosthesis. The prosthesis can thus be inserted with less force. Even though the prosthesis can now be secured with light blows, these blows must not be directed against the head or neck part.

The angles formed by the prosthesis and femur will be decided by anatomic conditions. The marrow space anchorage of the prosthesis can probably only be positioned in one way in the upper femur. Due to the direction of the longitudinal axis of the proximal femur, the prosthesis will form a ventral angle in the sagittal plane. This angle (α. in fig. 1) varies slightly between different individuals. According to Backman (1957) this angle is 8° with a standard deviation of ± 1.93°. In evaluating the direction of force, this angle must be taken into consideration.

The cervico-diaphyseal angle is normally about 126°. Backman (1957) has found its supplementary angle to be 54° with a standard deviation of ±5°. The angle formed by the prosthetic neck and the stem is 120°. As the stem follows the longitudinal axis of the proximal part of the femoral shaft, the angle formed by the prosthetic neck and the femoral shaft will be 120°, approximately.
The true cervico-diaphyseal angles is, however, a few degrees greater (page 35). The difference was only 1° for case 1 and 5° for case 2. The suitability of the selected neck-stem angle is questionable and an angle of 125° seems to be more advantageous. In evaluating the direction of forces on the prosthetic head, differences of the cervico-diaphyseal angle from the normal must be regarded.
Concerning the antetorsion angle, it is possible that the prosthesis will have the antetorsion angle which normally occurred in the joint previous to the operation. By measuring the unoperated side approximate information of anatomic conditions in the prosthetic hip-joint is obtained, but the degree of ante-torsion may vary considerably between the left and the right side. It is the only way, however, of getting an idea of earlier anatomic conditions in the case in question. For this reason the cervico-diaphyseal angle and the antetorsion angle for the sound hip-joint have been determined at the X-ray Department of the Ekman Hospital in Gothenburg in collaboration with Billing.
In collaboration with Backman at the Röntgenological Department in Lysekil the ante-torsion angle has been determined for the hip-joint with the measuring prosthesis. The antetorsion angle on the prosthetic side and on the opposite side seemed to agree well. Hence, no consideration has been given to eventual errors in the ante-torsion angle in determining the direction of forces acting on the prosthetic head. With regard to the cervico-diaphyseal angle, it has been assumed that originally this angle was of the same magnitude in the prosthetic hip as in the non-operated hip.
The values obtained from the two cases are presented in table 13.

	Prosthesis 1		Prosthesis 2	
	opposite side	prosthetic side	opposite side	prosthetic side
Cervico-diaphyseal angle	130°	120°	140°	120°
Antetorsion angle	0°	8°	38°	35°

Table 13. *Measured cervico-diaphysial and antetorsion angles. W=75 kp.*

Sterilization.

To prevent the occurrence of insulation faults, the prostheses must be sterilized in dry heat and the temperature should not exceed 125°. Sterilization was therefore carried out in a Waldén type autoclave with dried steam. In this autoclave the relative humidity will be max. 2 % and the sterilizing temperature was set at 115° C. The prostheses were sterilized for a quarter of an hour. After test-sterilization and bacterial cultivation the prosthesis showed no growth of bacteria.

Case selection.

In selecting persons on whom measuring prostheses are to be inserted, the following conditions will apply:

1. The patient's condition must be such that arthroplasty with a conventional prosthesis is required.
2. The patient must be willing to cooperate, and also be intelligent enough to ensure satisfactory cooperation.
3. Any changes in the cartilage of the acetabulum must not exist.
4. The other hip-joint must be perfectly sound, and a normal walking pattern must have existed before the actual condition.
5. The patient should not be too old.

Case 1 was a man, 51 years of age, who sustained a fracture of the head of the right femur in a motor accident. With the exception of an excoriation on his face, no other injuries could be detected. Two loose fragments had been split from the head of the femur. A rather large fragment was dislocated backwards and upwards, while a somewhat smaller fragment remained in the acetabulum. Otherwise, no changes could be found on the X-rays. The operation (Hirsch and Rydell) was performed 4 weeks after the accident with a posterior approach. The gluteus maximus was split lengthwise, musculus piriformis, obturator internus, obturator externus and quadratus femoris were cut, and the loose fragments removed. At the time of the operation the cartilage in the acetabulum showed no macroscopic changes. No signs of further injury to the bone structure could be found. The femoral neck was adjusted to fit the prosthesis which was secured to the marrow-space after a bed had first been made in the femur with a specially designed tool. The prosthesis was reduced into the acetabulum. The prosthesis ball fitted well in the acetabulum, but its diameter was 3 mm smaller than the femoral head on the other side. The joint capsule and the muscles were sutured and the cable pulled in a tunnel under the muscle fascia along the lateral-dorsal part of the femur reaching more than 2/3 the length of the femur. The contact cylinder was left under the fascia. The skin was sutured with catgut and steel wire.

The patient was confined to bed for four weeks, during which time exercises were performed. After four weeks the patient was allowed to get out of bed and then gradually to put weight on the operated leg. He received exercises in walking. Six months after the operation EMG tests were made giving results: "Normal EMG from gluteus medius, vastus medialis and lateralis and the adductor musles. The EMG obtained from the gluteus maximus had an atypical pattern, which would agree satisfactorily with the effect of the operation." (I. Petersén). As, by this time, the patient was walking without any noticeable limp, the time appeared to be suitable for the measuring tests. The patient's operated leg was then found to be 1.0 cm shorter than the sound leg. The patient preferred having a 0.5 cm thick inlay in the shoe of the prosthetic side. Because of the risk of infection involved by having a connection between the prosthesis in the hip-joint and the external environment, the examinations were terminated after six days. Since the female contact had to be brought out by surgery, the wound might have affected the walking ability of the patient, and therefore no examinations were started until two days after this operation. One exception to this was the examination made on the operation table during the awakening from the anaesthetic. The contact cylinder could be felt and by means of an incision the cylinder was brought out. Fascia and skin were sutured. The socket of the cylinder was removed and the first examination was carried out during his awakening from the anaesthetic. The socket was then replaced and the patient was permitted to walk from the operation table to his bed. This was done so as to make sure that the gauges gave the signals expected. The first day the patient had some discomfort from the wound, but after 24 hours it disappeared.

After the last examinations on the 6th day the teflon collar and the reinforced teflon strips which had been inserted between the leads were removed. The wires were then pulled one by one towards the aforedescribed steel cylinder and then pulled out. The wound healed quite normally.

Case II applied to a woman 56 years of age and in good health, who fell down from a table and fractured her left femoral neck. The patient was treated at home for 9 days and then sent to the hospital. The X-rays showed a medial femoral neck fracture in a severe varus position and a dorsal tilt. The femoral head was located practically under the neck. During the operation (Hirsch and Rydell), which was performed with a technique similar to that of the preceding case, a complete avascular head was found, but the acetabulum appeared to be normal. The diameter of the femoral head was exactly the same as for the prosthetic sphere.

The treatment after the arthroplasty was the same as in the first case.

Seven months after the operation and when the patient walked normally, EMG tests were made.
"From the gluteus medius and gluteus maximus on the left side no fibrillar action potentials were recorded. Voluntarily, a number of potentials are activated, and these are often clumsier in shape than normal and possibly of somewhat reduced quantity. Furthermore, it appeared to be more difficult to find active muscle tissue than in corresponding examination on the other side. The picture is not quite normal and must possibly be interpreted as the result of some mechanical muscular influence. At any rate, there are no signs of any neurogenic injury. The EMG from the adductors of the left thigh was entirely normal, like EMG from rectus femoris." (I. Petersén)
In this case the examinations were terminated after 8 days.

III. Electronic walk-ways and film equipment

Forces acting on the head of the femur in walking should vary as to magnitude as well as direction in a double step. Since the strain in the hip-joint is great during the stance-phase, and walking is perhaps the most frequent cause of strain of any appreciable magnitude in this region, measuring the force on the hip-joint in walking is an important part of the investigation.

The force which acts on the hip-joint in walking consists of one component determined by the superimposed body weight and balancing muscular forces and one component determined by the acceleration of the centre of gravity in the superimposed body mass. The last component may affect the first one, which has its peak approximately in the middle of the stance-phase, by making possible a balancing of the trunk without having to utilise the musculature fully. It is even possible that in the stance-phase the centre of gravity is shifted toward the hip-joint so that other load conditions apply than for one-leg support (Rossi 1963).

In the analysis of forces acting on the hip joint in walking, it is necessary to correlate the forces to the different phases of double steps: stance and swing-phase, heel-strike, toe-off, and double-support.

In walking, forces arise between the foot and the floor, which are generated from body-weight and acceleration of the centre of gravity of the body. The components of floor reactions in walking may be determined or recorded with suitable equipment, generally with so-called force-plates.

If recording with force-plates, one for the right foot and another for the left, is carried out concomitantly with the recording of forces acting on the hip-joint, the latter could be correlated to a double step's different phases. When evaluating the results, one must take into consideration that the vertical and horizontal force components in the hip-joint as described in part II are recorded to a prosthetically fixed coordinate system, while the forces which are recorded with force-plates are referred to a horizontally fixed coordinate system.

The vertical force component of floor reactions during walking is dominant. If this is also the case for force components acting on the hip-joint in walking, which is probable, the force curve which is recorded from the hip-joint should have the same appearance as the course of the vertical force recorded with a force-plate.

Construction of the electronic floor

Floor reactions during walking are often determined by means of force-plates. In the chapter of references, there is a brief outline on the development from pneumatic to electronic force-plates. With the electronic force-plates in use at present, four force components are usually recorded, the vertical force, fore and aft shear, the lateral horizontal force and the torsional moment around a vertical axis. There is one disadvantage in measuring with force-plates, namely that consecutive steps must be registered with a number of plates and the distance between them must be adjusted to the stride length of the person who is being tested. This becomes expensive and complicated and even if the force-plates are placed individually, one cannot preclude that the plate might have some effect on the walk when the foot is placed on it. In the present work, as readings are to be used mainly to correlate the forces acting on the hip-joint to various phases in a double step, an electronic-floor has been designed which allows measurement of consecutive steps at the same time as the fore and aft shear and the vertical component of the floor reactions are recorded.

Two identical 5-metre long walking-plates, one for the right foot and one for the left, have been constructed (fig. 22). Each walking-plate consists of two U-beams, joined by flat bars, force transducers and stabilizing equipment. The U-beams, with their hollow faces together form, with the flat bars, a framework which gives good stability and reduces the risk of deflection under load. The upper U-beam, the plane surface of which is the footpath, is 20 cm wide, while the lower one is 8 cm wide. The structure is made of light metal and each force-plate weighs about 30 kg. In

Fig. 22. *An outline drawing of the electronic walk-ways.*

view of the unsuitability of walking on a slippery metal surface, a thin layer of wadding and nylon fabric was glued on to the smooth surface of the upper U-beam.

Each walking-plate is suspended, via force transducers, from two inverted U-supports set in the concrete floor, and these run through the walking-plate's framework. The structure is suspended from the supports by way of the force transducers, under a vertical load, a tensile force arises in the transducers and the force may be registered in a suitable manner.

The fore and aft shear is recorded by means of a transducer connected to one of the ends of each of the upper U-beams and to a stand fixed to the floor with bolts.

In order to avoid exposing the force transducers to torsion or forces other than those which are to be registered, the upper U-beam is restrained in three places along its outer length to brackets screwed to the floor. In its upper part, each bracket is joined to the force-plate's upper U-beam by means of steel rods running perpendicularly to the U-beam provided with a universal joint. The universal joint allows free movement vertically and horizontally and does not influence the recordings of measured forces.

The surrounding floor was raised so that the smooth walking surfaces of the upper U-beams are at floor level.

Normally, during walking, a minor overlapping may occur between the feet in relation to the body's sagittal plane. To prevent this from influencing the measurements, the two force-plates are 2 cm apart.

The components of the resultant force were measured with Bofors KRK-1 force transducers. The transmitting principle is based on the deformation of a ring which is loaded at two diametrically opposite points. By this method both tension and compressive strains along the periphery of the ring are obtained, and these are transformed into an electrical signal by means of strain gauges.

The force to be measured is transmitted from two end blocks to the ring through flexible tongues, giving the end blocks a certain freedom of movement sideways — an advantage in measuring force components in two directions. Owing to the force transducers' symmetrical design, they are practically insensitive to bending movements at the end blocks and, therefore, need not be exactly parallel.

For calibration of the force transducers, external calibration resistors connected between two bridge corners are used. An off-balance signal is obtained in this way, corresponding to a previously determined force value. The transducers have a spring constant K=1.3 F nom $^{3}/_{2}$ kp per mm: F nom=mark load. The transducers have a linearity deviation which

is less than 0.2 % at full deflection and accuracy exceeds 1 %. The bridge resistance is 350 ohm and a maximum bridge current of 50 mA is permissible. The permissible range of temperature for measurement lies between —40° and +70° C.
Force transducers with a measuring effective range of ± 200 kp were chosen for recording vertical forces, and transducers with a range of measurement of ± 100 kp for recording horizontal forces.
For feeding, balancing and calibrating the force transducers, a balancing and calibrating unit, type BOFORS BK-3, was used.
The force transducers recording vertical forces, are connected in series, the same reaction is obtained irrespective of the point at which the beam is loaded.
Measurement signals are registered by means of a 12-channel Honeywell Visicorder, type 906 T. Miniature galvanometers M 100—120 A have been used.

Measurements with the electronic floor

When measurements are carried out with the electronic floor described above, two force curves for each foot are obtained, one for the horizontal and one for the vertical force component during the stance-phase. The curves look like those which have been obtained with various types of force-plates. The shape of the curves is constant for different steps and between different tests persons. The size of the readings and the time intervals between various points on the curves vary. In order to facilitate the interpretation and analysis of the curves and their variations, certain points have been marked with letters (fig. 23). For the vertical force curve, F, heel-strike is marked with *a* and corresponds to the point on the curve-

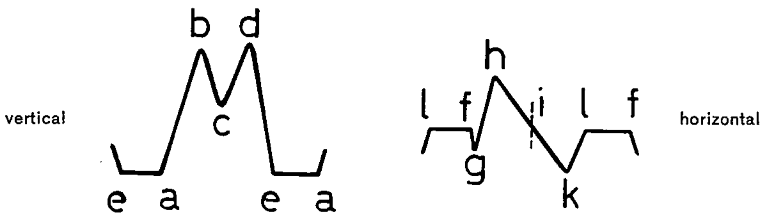

Fig. 23. *Shape of the vertical and horizontal force curves recorded by the electronic walk-ways. Points refered to in the text are indicated with letters.*

where the force sequence begins. The course of the curve is marked by two maximum points with an intermediate minimum point. The first maximum point is marked with *b,* the second with *d* and the intermediate force minimum with *c.* Toe-off, that is to say the point where the foot loses contact with the force-plate, is marked with *e;* thus *a—e* represents the stance-phase and *e—a* the swing-phase. Naturally, during the swing-phase no force is exerted between the foot and supporting surface. By simultaneous recording of the left and right foot, the times for double-support, i. e. the time when both feet rest on the ground, may be determined. The horizontal force curve H, has a somewhat different appearence. At heel-strike, there is a force peak aimed in the direction of walking. This force does not occur in all the steps. After this there is a new force peak, now against the direction of walking, after which comes a force peak in the direction of walking and the force returns to 0 at toe-off. The point on the horizontal curve which corresponds to the heel's contact with the force-plate is marked with *f,* the first force peak with *g,* the second with *h,* the third with *k* and finally toe-off with *l.* The point on the curve between *h* and *k,* when the force is *0,* is marked with *i.* The stance-phase in respect of the horizontal force curve corresponds to *f—l.* Due to the relatively high frequency of the force *g,* this force is difficult to assess and has not been subjected to a closer analysis. In measurements between different test persons, or on different occasions on the same person, curves with similar appearances are obtained. In slow walking, i. e. in walking speeds of less than 1 m/sec., the vertical curve may look slightly different. The change consists of an additional force minimum between the two peaks *b* and *d.* Sometimes there is an irregularity, usually between *a* and *b,* in the sequence of the curve, but the nature of this has not been subjected to further analysis (fig. 24).
In order to determine variations in the floor reactions in different walking

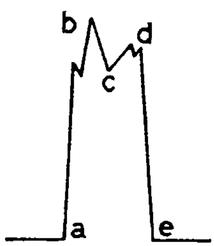

Fig. 24. *Irregularities occasionally present in the recordings from the walk-ways.*

speeds various tests have been carried out. Firstly, the curves for 10 men and 10 women were compared. The women were tested wearing shoes with low heels, medium heels and high heels. Comparing the curves for the male and female test persons a slight difference could not be ruled out in the cases where the women had worn high heels. The impression was that the point *d* was somewhat less in walking with high heels. This was not the subject of further analysis and in this work no data were obtained for walking in high heels. The criterion by which the persons tested could be considered to have a normal walk was that they had not had any illness or deformity in the lower extremities, that they themselves thought they had a normal walk and that, objectively, nothing noteworthy could be observed in regard to the way they walked. On closer comparison between the persons tested, only curves from tests where ordinary low-heeled shoes had been worn were used. Before the measurements were undertaken, the subjects were required to make trial walks several times in order to become relaxed. In the recording, a travel of 21 mm/sec. was used and the time marking was carried out by means of perpendicular lines appearing on the paper at 1/10 sec. intervals. In order to read at time to a hundredths of a second accurately, a vernier graded in tenths of seconds and the slide graded in 9/10 was manufactured. Checks have shown that readings can be made with an accuracy of $\pm$ 1/200 sec. Calibration of the walking-plates took place with the aid of a calibration resistor. A certain reading on the paper corresponded to a specific force. By varying the bridge current, the vertical force was adjusted so that the distance between each horizontal line on the paper corresponded to a load of 4 kp, while for the horizontal force each line was equivalent to a force of 2 kp. A static calibration was then carried out by applying a known load. A reading accuracy of $\pm$ 0.2 kp was obtained for the vertical force, while the accuracy for the horizontal force readings was $\pm$ 0.1 kp. The force transmitters give reliable answers to very rapid dynamic sequences and the reflecting galvanometers for frequencies up to 60 cps. The factor which determines the values' reliability is the walking-plates' natural frequency. This lies at about 50 cps, but under load the mass increases and the frequency will probably go down still further. The registered forces, in all cases except the force g for the horizontal curve, have a frequency which falls short of 6 cps, and a sufficient margin should exist. The force g as measured by the force-plate has a frequency of 25 cps, making the measurement values for this force unreliable.

Walking speed was determined with the help of a stopwatch. The test persons were permitted to use the stride length which seemed natural to them at different walking speeds.

In the tests on the walking-plates, four successive stance phases were measured for the one leg and three for the other. The first and last steps on the walking-plates showed large deviations from each other. The force *d* was considerably greater than *b* for the first step, while for the last step the force *b* showed an excess compared to the force *d*. As each end of the structure is situated at a distance of only one metre from the walls, this has been construed as a sign of acceleration in regard to the first step and retardation in regard to the last. In the measurements, therefore, no values for the first and last steps were used.
In order to find out the dissimilarities in the force sequence between two successive steps, five different persons were tested on ten different occasions. There was no appreciable difference as regards the time intervals and the forces for the vertical components. The forces *h* and *k* for the horizontal component did, however, show *a* systematic variation. The force *h* and *k* was almost always somewhat greater for the second of the analysed steps. However, this has been considered insignificant for the correlation of forces acting on the hip joint to forces between the foot and the floor and there has been no attempt to find an explanation for this.
Ten test persons were asked to walk slowly, at a medium pace and quickly, and registration was carried out with the electronic-floor. No measurement of the stride length was undertaken; the test persons simply walked with a stride which felt natural to them. The data obtained in these tests were handed over for statistical processing to Mr. Carlström, Lecturer at the Statistical Institute, Gothenburg. The following report was received.
The statistical analysis concerns the study of the relation between walking speed and forces (force functions) versus time intervals.
The analysis of the association between forces and body weight demonstrates that the relation is linear and goes through the origin of coordinates. The forces have, therefore, been indicated in percentage of the body weight in the analyses. For all y-variables (= forces/body weight, time intervals), the following has been calculated:

$$\bar{y_x} = a + bx$$

where x denotes walking speed.
This has been carried out for both left and right observations, after which they have been compared in order to find possible differences between them. If no difference could be observed, the two lines have been weighted together to one line.
There is a significant, linear relation to walking speed in all the cases (except for g) and in no case is it possible to detect any significant difference between the right and left functions.

Registration on film

If forces acting on the hip-joint in walking are required in relation to the coordinates of the body, account must be taken of the position of the prosthesis in the acetabulum during e. g. heel-strike, toe-off. The normal values for these angles have been determined by Barry (1952). In order to see whether the patients with the measuring prostheses moved the legs in a normal way, a registration on film was carried out at the same time as the measurements with the electronic floor took place.

A track for the film equipment was constructed parallel to walking-plates. The track consists of two steel tubes, which are firmly moored to the concrete floor through supports and steel pillars. The camera, lighting device and a synchronizing clock are mounted on a trolley. This trolley runs on wheels with roller-bearings between the iron tubes and has practically vibration-free movements, which gives a high degree of sharpness in the film.

A 16 mm Paillard film camera with wide-angle lens Switar 1:1.6, f = 10 mm is used for filming. The film was exposed at the rate of 64 pictures/sec. The trolley with film equipment is propelled by hand at the same speed as the test person is walking (fig. 25).

For projection of the film, a projector which allows the feeding of one frame at a time is used. To facilitate measuring, the film is projected on to the back of a drawing board. In the middle of the drawing board top, a rectangle measuring 66×45 cm is removed and in its place a sheet of ground glass is inserted, on to which the film is projected. A tracing paper is placed over the ground glass sheet and on this paper the axes of the extremities may be drawn and angles determined.

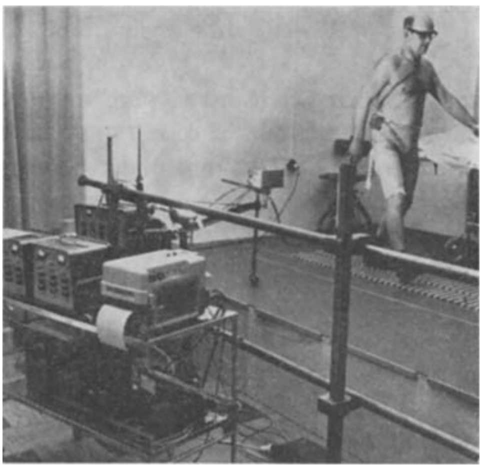

Fig. 25. *Case 1 walking on the electronic floor. The film track and the film equipment can also be seen.*

Fig. 26. *The synchronizing clock allowing a film frame to be connected to a certain point on the recorded curves.*

In the case of determining the longitudinal axes of the femur, leg and foot, difficulties may arise. The way to draw the axes of thigh and leg with the least faults may be debated. The method which came into use does not make any claims on being exact. In the present work consideration has been taken only to the thigh's longitudinal axis in relation to the vertical line. The thigh's direction was determined on the lateral side by marking the greater trochanter and the lateral femoral epicondyle and on the medial side by marking the mid-point on the most proximal part of the leg which could be seen, and the medial femoral epicondyle on the medial side.

One obvious disadvantage which is inherent in registration on film is that from the pictures, it is not possible to decide precisely at what moment the foot touches or leaves the ground. Therefore a synchronizing clock was designed (fig. 26) which marks the walking-plates' recording paper and whose dial may be read off in the film frame. In this manner a frame may be precisely synchronized to a certain point on the curves recorded by the electronic floor.

Results

All of the following mean values are presented with their standard deviations.

Although the subjects in which the measuring prostheses were inserted were selected with regard to age and normal anatomical conditions of the acetabulum, the recordings, when done were naturally not under physiological conditions, because a metal prosthesis never is equivalent to the parts it replaces and the surgical approach influences the function of the joint. Differences of angles and dimensions between the prosthesis and the part it replaces may give changes regarding lever arms and the gait may be more or less affected.

The gait of the two cases with the measuring prostheses, appeared to be normal, but a closer analysis showed discrepancies. The force-plates and simultaneous film recordings indicated that differences were present when compared to normal people. The angle v that the longitudinal axis of the thigh in the sagittal plane forms with the vertical line was measured during the heel-strike when the femur forms its greatest angle with the vertical. Simultaneously the other leg, which has not yet left the ground, forms its greatest angle in the sagittal plane with the vertical line and this angle called u was also determined (fig. 27). When walking is abnormal these angles v and u are changed.

The curve of the vertical component of floor reactions recorded by the force-plates is also influenced if the gait is changed. Pilot-tests performed on patients with abnormal gait showed that there is a great difference in the ratio of duration of stance-phase to swing-phase between the left and the right side. This difference appears even in slight gait disturbances. The angles v and u, and the stance-swing ratio were determined for the two prostheses cases and for normal persons and the results compared. The angle formed by the longitudinal axis of the femur and the vertical line in the sagittal plane was determined by means of the film frame, which corresponded to heel-strike for the prosthetic leg and the opposite leg, respectively. This frame was projected on the ground glass of the drawing board top. The longitudinal axis of the femur was ascertained by the method described previously. Even if the method has some shortcomings with regard to accuracy of angles and axes it is useful for comparison of cases.

Account has not been taken to the length of stride. The subjects were told to walk in a natural way. Regarding the cadence the subjects walked first slowly with a speed of about 0.8 m/sec. and then faster with about 1.2 m/sec.

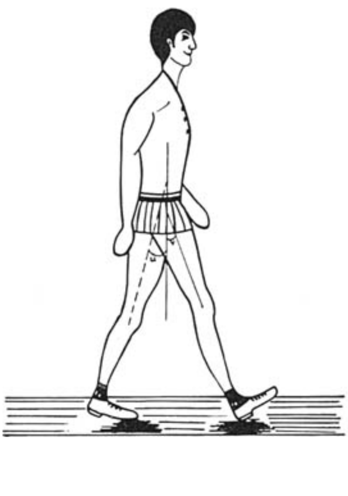

Fig. 27. *At heel-strike the fore leg forms an angle* v *with a vertical line and simultanously the hind leg forms an angle* u.

To determine the accuracy, with which angles can be determined with this method, known angles, which were drawn were filmed and the accuracy was found to be $\pm 1^{\circ}$. Greater difficulties arise when the angle is determined between the longitudinal axis of the femur and the vertical line. Tests were therefore performed with legs, which were filmed bent at known angle. The angles were determined with an accuracy of $\pm 2^{\circ}$. Berry (1952) has determined the angles v and u and found the angles to be normally 24° and 25°, respectively, in level walking. In order to compare these angles to the ones obtained in this investigation the angles v and u were determined for 14 persons with a normal gait. For each person six readings for every angle were made, three at a cadence of about 0.8 m/sec. and three of about 1.2 m/sec. The individual difference did not exceed 3°. The angle v had a mean value of $24.8^{\circ} \pm 2.1$ at heel-strike for the right leg and the angle u was for this phase 21.7 ± 3.3. At heel-strike for the left leg the angle $v = 24.2^{\circ} \pm 2.5$ and $u = 20.0^{\circ} \pm 2.2$. For the "case of prosthesis 1" the following values for angle v and u were recorded. At heel-strike for the prosthetic leg $v = 23.1^{\circ} \pm 1.8$ and $u = 10.0^{\circ} \pm 1.8$. At heel-strike for the opposite leg $v = 24.3^{\circ} \pm 1.6$ and $u = 12.4^{\circ} \pm 2.8$. For the "case of prosthesis 2" the following values were obtained. At heel-strike for the prosthetic leg $v = 21.7^{\circ} \pm 1.6$ and $u = 14.7^{\circ} \pm 2.1$ and at heel-strike for the opposite side $v = 22.3^{\circ} \pm 0.89$ and $u = 19.3^{\circ} \pm 2.06$. The results show that the angle u is less than normal in the prosthetic cases, and this is most pronounced when the prosthetic leg is at heel-strike.

In determination of the direction of the force acting on the prosthetic head the magnitude of the angles v and u must be taken into account as well as the fact that the longitudinal axis of proximal femur forms a dorsally opening angle with the ideal axis of the femur.

An abnormal gait also gives rise to changes from the normal in the curve of the vertical component of floor reactions. As mentioned earlier, pilot-tests showed that even in slight disturbances of gait, a difference between the left and right side will occur in the stance-swing ratio. The stance-swing ratio varies with the walking speed but no difference between left and right leg exists normally.

The curve of the vertical component recorded by the electronic floor showed for the "case of prosthesis 1" marked differences from normal. The most obvious difference was that the curves had irregularities rather frequently. Normally such irregularities may occur in the curve of the vertical component of floor reactions but not very frequently. The cause of those irregularities is obscure. One explanation may be that the prosthesis had lost its grip in the femur, but no indication for this appeared on the x-rays. As the irregularities also were pronounced in the curves recorded by the force-plates from the unoperated side it is not likely that loosening of the prosthesis is the cause of these irregularities. Another explanation may be that the diameter of the prosthetic head is 3 mm less than that of the femoral head of the opposite side, and there is a possibility of slackness in the joint. At increasing speed the irregularities decreased and at fast walking they were almost absent.

The difference of the stance-swing ratio between the prosthetic and the opposite side was great in case 1 at slow walking. At the walking speed of 0.7 m/sec. the stance-swing ratio was for the prosthetic leg 1.60 and for the opposite leg 2.57. At walking speed of 1.1 m/sec. the stance-swing ratio was 1.43 for the prosthetic side and 1.96 for the opposite side and with a walking speed of 1.3 m/sec., the ratio was 1.47 and 1.54, respectively. Differences of the ratio between two consecutive steps for the same side never exceeded 0.1. In a population of 31 normal persons, the ratio between stance-swing ratio on one side to stance-swing ratio on the other side was 1.06 ± 0.07.

For the case 2 the curve of the vertical component of floor reaction showed no irregularities and no difference between the prosthetic side and the other side appeared regarding the stance-swing ratio. The quotient of the stance-swing ratio of the good and the prosthetic side was in 17 observations 1.04 ± 0.07.

IV. Intravital measurements and results

The force acting on the prosthetic head has been determined as to its magnitude and direction under static and dynamic conditions.
Recordings were made for one-leg support, flexion, extension, abduction and while walking. In case 2 measurements were also performed while running. The recordings in flexion, extension and abduction were made both with and without movement.
The coefficent of friction between the metal surface of the prosthetic head and the cartilage of acetabulum has also been measured.
The results are valid for the prostheses only, but if factors mentioned on pages 57—58 and 83—85 are considered they may give us information which could be applied to normal conditions.
All of the following mean values are presented with their standard deviations.
The unit of force used in this work is kilopond (kp). 1 kp is defined as the force which gives the mass 1 kilogram an acceleration of 9.80665 m/sec.^{2}

Flexion

The tests were made with the patients in the supine position. A protractor with its centre situated at the top of the greater trochanter was used for determination of the amount of flexion. The patients were told to lift the leg with the knee straight to a point equal to 30°, 60° and 90° of flexion, respectively. During the tests it was difficult for the patients to keep the knee fully extended during hip flexion. The force acting on the

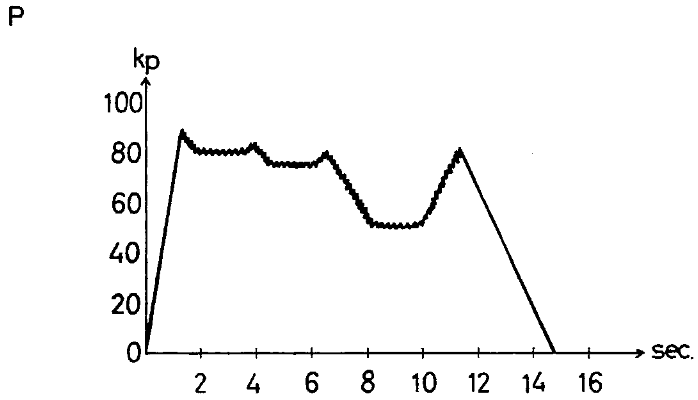

Fig. 28. *Recordings from flexion tests of case 1, when maintaining a certain degree of flexion the acting forces varied somewhat in magnitude.*

prosthetic head was always greater when the leg was moved. The differences between dynamic and static conditions were not great, except for the first 10° of flexion.

Case 1

Each test was performed three times. Between the tests there was an interval to allow the galvanometers to return to zero.
During the tests no recordings were obtained from section 1. When the hip maintained at a certain degree of flexion the curves showed that the acting forces varied in magnitude (fig. 28).
The amount of flexion in the knee joint during the tests varied about 10°—15°. No attempt was made to support the knee joint by means of splints or elastic bandages. The recordings obtained with prosthesis 1 can be seen in Table 14.

Amount of flexion	P_y kp			P_z kp			P_x kp			P kp		
30°	28	23	24	45	42	41	61	62	64	81	78	79
60°	33	29	27	47	44	41	52	51	54	78	73	73
90°	20	17	19	30	30	30	35	27	38	50	44	52
moving 0°—10°	30	33	30	42	45	49	70	74	78	87	92	97
moving 10°— 0°	37	41	40	38	38	35	68	64	66	86	84	85

	$\frac{P}{W}$			α°			γ°		
30°	1.08	1.04	1.06	24	20	21	59	61	60
60°	1.03	0.98	0.97	32	30	27	55	57	57
90°	0.67	0.58	0.69	29	32	27	57	61	58
moving 0°—10°	1.16	1.23	1.29	23	25	21	55	53	59
moving 10°— 0°	1.14	1.13	1.29	29	33	31	46	42	41

Table 14. *Case 1. Flexion of the prosthetic leg. W=75 kp (body-weight).*

Case 2

The tests were performed in the same way as in Case 1. From section 1 no recordings of importance occurred regarding M_{1z}. The readings for M_{1y}, however, were not negligible. These readings were negative and rather small at the beginning of flexion, but increased with flexion and were great at 90° of flexion. Account has further been taken of M_{1y} in evaluation of the force P_z. No irregularities when the leg was kept flexed were found, and the galvanometers returned quickly to their zero-points.
As in Case 1, there were difficulties in keeping the knee extended. In two tests, however, flexion was performed with the knee straight.

The results are presented in Table 15. In Table 16 the results of the two tests, when flexion was performed with the knee fully extended, can be seen.

Amount of flexion	P_y kp			P_z kp			P_x kp			P kp		
30°	21	20	20	12	13	12	42	36	36	47	43	43
60°	19	16	19	16	14	13	36	36	36	44	42	43
90°	12	11	12	16	14	14	30	30	33	36	35	38
moving 0°—10°	20	22	20	13	14	16	54	45	45	59	52	52
moving 10°— 0°	21	22	21	12	14	12	36	36	36	44	45	44

	$\frac{P}{W}$			α°			γ°		
30°	1.07	0.98	0.97	27	28	28	30	34	32
60°	0.99	0.95	0.97	28	25	28	40	39	34
90°	0.82	0.79	0.86	21	21	20	55	51	49
moving 0°—10°	1.35	1.18	1.18	21	26	24	33	34	37
moving 10°— 0°	0.99	1.01	0.99	30	32	30	30	32	30

Table 15. *Case 2. Flexion of the prosthetic leg. W=45 kp (body-weight).*

Amount of flexion	P_y kp		P_z kp		P_x kp		P kp	
30°	24	23	17	17	55	58	62	64
60°	21	17	18	18	49	58	56	62
90°	11	12	14	15	24	27	30	33
moving 0°—10°	25	25	14	14	85	82	90	87
moving 10°— 0°	23	23	17	17	36	49	46	56

	$\frac{P}{W}$		α°		γ°	
30°	1.40	1.46	24	22	35	36
60°	1.27	1.42	23	16	40	47
90°	0.69	0.75	25	24	52	50
moving 0°—10°	2.03	1.97	16	17	29	30
moving 10°— 0°	1.05	1.27	33	25	36	36

Table 16. *Case 2. Flexion of the hip with the knee fully extended. Prosthetic side. W=45 kp (body-weight).*

Flexion of the opposite leg. Case 1 and 2.

Forces acting on the prosthetic head were also recorded in the flexion of the opposite leg. The readings from M_{1y} were positive in the case of prosthesis 2, and negligible in prosthesis 1. The recordings from M_{1z} were negligible in both cases. Table 17 and 18 show the results for cases 1 and 2, respectively.

Amount of flexion	P_y kp			P_z kp			P_x kp			P kp		
45°	16	16	22	7	6	7	25	26	38	30	31	45
90°	10	14	13	2	2	2	12	12	11	16	19	17

	$\frac{P}{W}$			$\alpha°$			$\gamma°$		
45°	0.40	0.41	0.59	33	31	30	24	22	19
90°	0.22	0.25	0.23	40	49	50	13	9	10

Table 17. *Case 1. Flexion of the hip. Opposite side. W=75 kp (body-weight).*

Amount of flexion	P_y kp			P_z kp			P_x kp			P kp			$\frac{P}{W}$		
45°	21	26	22	4	4	3	13	16	13	25	31	26	0.57	0.70	0.58
90°	13	13	15	6	5	3	12	12	12	19	18	19	0.42	0.40	0.44

Table 18. *Case 2. Flexion of the opposite leg. W=45 kp (body-weight).*

Discussion

The irregularities occurring in Case 1 could partly be of the same origin as the irregularities in the curves during walking.

In Table 19 the mean values of angles and the ratio $\frac{P}{W}$ of the two cases in flexion on the prosthetic side are shown. Although there are differences between the two cases, some conclusions may be drawn. The force P_z is always acting from the ventral side running in a dorsal direction. The magnitude of the force P exceeds body weight when flexion is between 0° and 60°.

Flexion of the hip gives rise to a force in the same side of about body weight if the knee is not fully extended, but with a straight knee the force can reach the level of twice the body-weight. In the opposite side a force of about 1/2 times the body weight will act.

Amount of flexion	Case 1 $\frac{P}{W}$	α°	γ°	Case 2 $\frac{P}{W}$	α°	γ°	Case 2 straight knee $\frac{P}{W}$	α°	γ°
30°	1.06	22	60	1.02	28	32	1.43	23	35
60°	0.99	30	56	0.97	27	38	1.34	20	43
90°	0.65	29	58	0.82	21	52	0.72	25	51
moving 0°—10°	1.22	23	55	1.23	23	35	2.00	18	57
moving 10°— 0°	1.13	31	43	1.00	31	31	1.04	28	60

Table 19. *Case 1 and 2. Approximate values of* $\frac{P}{W}$ *and angles in flexion of the prosthetic leg. W case 1 = 75 kp. W case 2 = 45 kp.*

Extension

The tests were performed with the subjects in prone position and with the leg extended. Maximal extension was desired, but in each test an additional rotation of the pelvis occurred, more pronounced in Case 2. Normally, the amount of extension is 15°, but it is very difficult to perform maximal extension without rotation of the pelvis. As in the prone position, a small force was acting on the prosthetic head, the zero lines had to be taken from a relaxed supine position. The force acting in the prone position gave rise to readings from the curves of M_{2y} and M_{2z}. Readings from M_{2y} were positive and for M_{2z} negative. The force P_y varied in different tests between 1 — 3.7 kp and acted in a cranial direction while the force P_z varied between 1 — 3 kp, acting in a dorsal direction. The force acting on the prosthetic head in a relaxed prone position might be explained by a moment developed by a force acting on the knee from the underlying surface, or due to muscle forces.

Case 1

Each test was repeated three times. The patient was told to extend the leg fully. Rotation of the pelvis occurred in every test. In extension no readings were recorded from section 1, and the forces obtained can be seen in Table 20.

P_y kp			P_z kp			P_x kp			P kp		
50	52	52	17	17	14	78	86	84	94	102	100

$\frac{P}{W}$			α°			γ°		
1.25	1.35	1.33	33	30	32	19	18	15

Table 20. *Case 1. Maximal extension of the prosthetic side. W=75 kp.*

Case 2
The tests were performed in the same way as in case 1. In extension, account had to be taken of readings from the curves of M_{1y} and M_{1z}. The rotation of pelvis during extension was more pronounced than for case 1. The results are shown in table 21.

P_y kp			P_z kp			P_x kp			P kp		
41	42	41	9	8	7	83	83	83	92	93	93

$\frac{P}{W}$			α°			γ°		
2.10	2.11	2.10	26	27	27	13	10	10

Table 21. *Case 2. Maximal extension of the prosthetic hip W=45 kp (body-weight).*

Extension in prone position with the opposite leg
This test was only performed in Case 2. Even in this test a rotation of the pelvis occurred. The results are presented in table 22.

P_y kp	P_z kp	P_x kp	P kp	$\frac{P}{W}$	α°	γ°
39	17	56	70	1.56	35	24
39	18	56	70	1.56	35	25
39	19	59	73	1.62	33	27

Table 22. *Case 2. Extension of the hip. Opposite side. W=45 kp (body-weight).*

Discussion
Forces acting on the hip-joint during extension and rotation of the pelvis are rather great. In case 1 the force exceeded body-weight by about one-third, and in case 2 the forces were more than twice that of the body-weight.
Extension of the opposite leg gives rise to a surprisingly great force exceeding body-weight by more than 50 %.

Abduction

The tests were performed with the patients in supine position lying on the smooth surface of a plate. This plate was graduated to facilitate the determination of the amount of abduction. The subjects were told to abduct their legs to 30°. Moving of the pelvis was not allowed. The maximal values recorded during abduction were determined.

Forces acting on the prosthetic head when abducting the opposite leg were also determined.

Case 1

In these tests the patient lying on the smooth surface was told to abduct the leg without bending the hip and knee. Table 23 shows the results obtained during abduction.

Under the same conditions as for abduction of the prosthetic leg, tests were performed with abduction of the opposite leg. The results are seen in table 24.

Abduction	P_y kp			P_z kp			P_x kp			P kp		
0°—30°	26	18	16	23	14	14	35	27	37	49	35	43

	$\frac{P}{W}$			α°			γ°		
0°—30°	0.66	0.46	0.57	37	33	23	42	39	43

Table 23. *Case 1. Abduction of the prosthetic leg. W=75 kp (body-weight).*

Abduction	P_y kp			P_z kp			P_x kp			P kp		
0°—30°	12	12	12	6	6	6	13	13	12	19	19	18

	$\frac{P}{W}$			α°			γ°		
0°—30°	0.25	0.25	0.24	43	43	45	27	27	27

Table 24. *Case 1. Abduction of the opposite leg. W=75 kp (body -weight).*

Case 2

As some degree of flexion during abduction in the tests of Case 1 could not be excluded, the tests were this time performed in a different way. The subject was lying on the same plate but had a specially made ball-bearing support between the heel and the plate. In this way the need of using the flexor muscles when abducting the leg was decreased. The results can be seen in table 25.

The abduction of the opposite leg was performed with the same technique and the forces acting on the prosthetic head were recorded. The results are seen in table 26.

Abduction	P_y kp			P_z kp			P_x kp			P kp		
0°—30°	5	6	4	5	7	5	30	33	27	31	34	28

	$\frac{P}{W}$			α°			γ°		
0°—30°	0.69	0.77	0.62	8	10	8	51	51	50

Table 25. *Case 2. Abduction of the prosthetic leg. W=45 kp (body-weight).*

Abduction	P_y kp		P_z kp		P_x kp		P kp	
0°—30°	4	2	2	3	6	6	8	7

	$\frac{P}{W}$		α°		γ°	
0°—30°	0.17	0.16	33	18	32	59

Table 26. *Case 2. Abduction of the opposite leg. W=45 kp (body-weight).*

Discussion

The magnitude of the force P in relation to body-weight is relatively equal in the two cases. There is a marked difference in the values of the angle α and a small difference of the angle γ between the two cases. In table 27 approximate values for the angles and the ratio $\frac{P}{W}$ of the two cases in abducting the prosthetic side are presented.

Tests in adduction were excluded for technical reasons. It was impossible to perform adduction without either flexing the leg or abducting the opposite leg.

Case 1			Case 2		
$\frac{P}{W}$	α°	γ°	$\frac{P}{W}$	α°	γ°
0.56	31	41	0.69	9	51

Table 27. *Case 1 and 2. Approximate values of $\frac{P}{W}$ and angles in abduction of the prosthetic leg.*
W case 1=75 kp. W case 2=45 kp (body-weights).

Traction

To determine the effect of traction in the hip-joint the patient in case 2 was placed in the supine position and traction applied to the leg. Tests were performed with the knee in flexion and extension. The load was applied gradually from zero to 24 kp. No force was recorded even after application of a load of 24 kp for 3 hours. If a load was applied suddenly, a force was shown to be present for approximately 0.1 sec.

Sitting

These tests were performed while sitting in a chair with the hips and knees flexed to 90° and with and without foot support. In case 1 no forces were recorded while sitting.

In case 2 small forces between 3—10 kp were present while sitting. Recordings were mainly obtained from P_z. P_x was always zero; P_y negative and hardly measurable.

One-leg support

Standing on the prosthetic leg

The patients were told to stand on the prosthetic leg. The opposite lower extremity was kept free from the floor by flexing the knee, but some flexion in the hip did occur. Both patients had some difficulty in maintaining this position, and in trying to maintain equilibrium they swayed to and fro. This was most pronounced in case 1, who when attempting to maintain his pelvis horizontal often curved his back. In case 2 the same problems arose but to a lesser degree. Swaying was not as marked, and no scoliosis occurred. Each time the desired position; a horizontal pelvis and no curvature of the spine was maintained for a period of some seconds; this was indicated on the recording paper.

Case 1

Readings were made from the tests, in which the patient remained in the desired position long enough for indication. Even when the patient seemed to be steady, oscillations of a rather high frequency occurred. Readings were taken in the parts of the curve where irregularities were least pronounced.

In section 1 recordings occurred for the two moments. M_{1y}, however, had its deflections around zero in such a way that they influenced the magnitude of the force P_z very little. The deflections for M_{1z} occurred in a similar way, but around a line positive to the zero line. In the determination of the force P_y account had therefore to be taken to M_{1z}. In table 28 the magnitude and direction of the forces acting on the prosthetic head, when standing on one leg, can be seen.

P_y kp	P_z kp	P_x kp	P kp	$\frac{P}{W}$	α°	γ°
108	15	135	173	2.3	39	8
112	13	130	172	2.3	41	6
110	12	135	175	2.3	39	6
113	11	125	169	2.2	42	6
113	16	130	173	2.3	41	8
110	15	134	174	2.3	39	8

Table 28. *Case 1. One leg support. Prosthetic leg. W=75 kp (body-weight).*

Case 2

When standing on the prosthetic leg the swaying was of lower frequency and magnitude than in case 1. The curves were therefore smoother and easier to interpret.

Account had to be taken of both M_{1z} and M_{1y} in section 1 in determination of the forces P_y and P_z, respectively. M_{1z} was mainly negative, but in some tests positive, while M_{1y} was always positive. Readings were made in the parts of the curve where the irregularities were smallest. Recorded forces and angles obtained are presented in table 29.

P_y kp	P_z kp	P_x kp	P kp	$\frac{P}{W}$	α°	γ°
67	10	101	120	2.7	34	9
68	11	106	120	2.8	33	9
68	12	106	125	2.8	33	10
66	13	101	120	2.7	33	11
70	13	106	125	2.7	33	11
67	13	113	129	2.9	31	11

Table 29. *Case 2. One-leg support. Prosthetic leg. W=45 kp (body-weight).*

Discussion

The force acting on the prosthetic head was on the average 2.3 times body-weight in case 1 and 2.8 times body-weight in case 2. The angle α was approximately 40° and 33°, angle γ 7° and 10°, respectively. As can be seen from tables 28 and 29, the differences in magnitude and angles are surprisingly small between different tests in the same case. There are differences, however, between the two cases, but regarding the directions of forces, the differences are small.

In case 1 irregularities in the curves produced difficulties with the readings. As noted, readings were taken from the smoothest parts of the curves, however; because of oscillations a mean value was read.
The oscillations were of lower frequency and magnitude in case 2, and smooth parts of the curves were easily found for readings.
As maximal forces occurred in swaying to maintain equilibrium in what seemed to be rather static conditions, it was believed that readings might be of interest when the oscillations were at their peaks.
In case 1 the force P was found to be about 190 kp or 2.5 times body-weight, if the recordings were evaluated for the maximal force in one-leg support. In case 2 the moment M_x was also recorded and the maximal force occurred when M_x changed its direction. The maximum force recorded in one leg-support, when standing on the prosthetic leg was about 140 kp or 3.1 times body-weight. One-leg support is often the subject of force analysis and it might be of interest to compare the magnitude of the forces obtained by the prosthesis with theoretical values. Analyses have therefore been carried out according to the laws of mechanics in a similar way to those described by Williams and Lissner (1962). This is a way to get as close as possible to the conditions existing in the operated hip-joint prior to the accident. Calculations were made both for the opposite and for the prosthetic side.
According to Williams and Lissner (1962) fig. 29, the direction of the force generated by the abductor muscles was assumed to form 71° with the horizontal plane on the normal side. Assuming further that the same conditions existed on both sides, the direction of the abductor pull on the prosthetic side was determined in the following way.

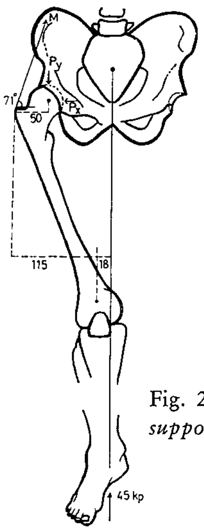

Fig. 29. *Determination of the force acting on the femoral head in one-leg support. Figures from case 2.*

Line-drawings were made of the pelvis and the two femurs from x-rays taken in one-leg support. The drawing of the prosthetic femur was placed in the opposite acetabulum with the shaft in the same direction as in one-leg support. A line was drawn from the greater trochanter to the point representing the intersection between the normal direction of the abductor pull and the iliac crest. This line was assumed to represent the direction of the abductor pull of the prosthetic hip. It should be understood that this method gives only approximate values.

Presuming the leg to be $^1/_6$ of the body-weight, the force P acting on the femoral head in case 1 was calculated on the non-operated side to be 150 kp or 2.0 times body-weight and on the prosthetic side to be 139 kp or 1.8 times body-weight. The force P forms an angle with the horizontal plane on the non-operated side of 78.8° and on the prosthetic side of 76.0°. According to the differences of the cervico-diaphyseal angles, the angle α between the force and the longitudinal axis of the neck is on the non-operated side 38.8° and on the prosthetic side 46.0°.

In case 2 the force P was found to be 105 kp on the non-operated side and 91 kp on the prosthetic side, which is 2.3 and 2.0 times body-weight, respectively. The angle formed by the force and the horizontal plane was on the non-operated side 77.7° and 75.0° on the prosthetic side and the angle α 27.7° and 45.0°, respectively.

The differences in magnitude of the recorded forces 2.3 and 2.8 times body-weight between the 2 cases are in accordance to the differences between the theoretical values. The recorded forces are higher than expected from theoretical calculations.

Regarding the direction of the force the theoretical calculations were only made for the frontal plane. The recorded angle α between the force and the cervical axis of the prosthesis is smaller than the theoretical angle, which is most pronounced in case 2. This can be due to either a more adducted femoral shaft during the tests or to muscle effects. If muscle forces are the cause, this might also explain the higher magnitude of the force. Theoretical calculations were made with the femur in a more adducted position. A slight increase of the magnitude of the force was obtained, but the direction was unchanged. This is in agreement with Innman (1947), who claims that the direction of the reactive force through the head of the femur is constant and, therefore, independent of the position of the pelvis.

Standing on the opposite leg

The patients were told to stand on the opposite leg in both cases. Maintaining equilibrium was easier compared to standing on the prosthetic leg. However, small oscillations occurred in case 1.

The non-supporting leg can be positioned in different ways, e.g. with flexion in the hip and knee joint and only with flexion in the knee joint, the hip in a neutral position or slightly in flexion, extension or abduction.

Case 1

Because of some difficulties in maintaining equilibrium in this case, the non-supporting leg was kept with the knee joint in flexion and the hip was noted to move slightly, involuntarily, around the neutral position.

As usual, the readings were made in smooth parts of the curves. The results can be seen in table 30.

P_y kp	P_z kp	P_x kp	P kp	$\frac{P}{W}$	α°	γ°
32	15	34	49	0.65	43	26
20	14	18	30	0.40	48	36
19	10	25	33	0.44	37	28

Table 30. *Case 1. Standing on the opposite leg, the prosthetic leg bent at the knee joint but around neutral position in the hip. W=75 kp (body-weight).*

Case 2

Difficulties in balancing were slight, equal to those seen in any patient. The non-supporting leg was kept with the knee joint and the hip joint in 90° and 45° of flexion, respectively. The curves appeared smooth with isolated peaks. Forces and angles obtained are shown in table 31.

P_y kp	P_z kp	P_x kp	P kp	$\frac{P}{W}$	α°	γ°
18	22	29	41	0.91	32	51
17	23	29	41	0.91	31	54
19	20	32	42	0.94	30	47
17	21	29	38	0.86	31	50

Table 31. *Case 2. Standing on the opposite leg, the prosthetic leg bent in hip and knee joint. W=45 kp (body-weight).*

Discussion

Tests for the same subject showed differences to be rather small, but test differences of the two patients are significant. As the tests were not performed in the same way in case 1 and 2, the results are not comparable. The different behavior of the two patients in standing on the opposite side was noticed after the evaluation of the readings, and it was then too late to perform more tests. The size of angle γ shows that the force acts in a more horizontal direction in case 2 according to the coordinates of the prosthesis, which is in agreement with the way the tests were performed. In case 2 tests were also performed standing on the opposite leg, swinging the prosthetic leg to and fro. The leg was swung from about 45° of flexion to full extension. A small rotation of the pelvis in full extension could not be excluded. M_{1z} and M_{1y} had to be taken into account in determining the forces P_y and P_z.

Readings were made at the turning points a and b. Point a corresponds to the instant when the leg in full extension changed direction of motion from dorsal to ventral, and point b to the instant when the leg in flexion changed direction from ventral to dorsal. The torsional moment around the x-axis M_x changed at point a from positive to negative, and at point b from negative to positive.

At point a, P_x and P_z were maximum, while P_y increased and reached its maximum about 2/10 of a second later. At point b, P_x and P_z had another maximum but P_y was at a minimum.

The forces and angles obtained in these tests can be seen in table 32.

	P_y kp	P_z kp	P_x kp	P kp	$\frac{P}{W}$	α°	γ°
a	28	3	27	39	0.87	45	7
b	6	20	20	29	0.65	16	68
a	23	4	27	36	0.80	40	9
b	3	19	20	28	0.62	9	81
a	26	5	22	35	0.78	50	11
b	3	21	22	31	0.69	8	81

Table 32. *Case 2. Standing on the opposite leg, swinging the prosthetic leg from full extension to 45° of flexion. Point a corresponds to the change of direction in full extension and b to the change of direction in maximal flexion (45°). W=45 kp (body-weight).*

Level walking

Forces acting on the prosthetic head during level walking were determined. To get information on loading conditions which affect the lower extremities and the transition between the stance- and swing-phase, recordings were made using the electronic walk-ways described in part III. At the same time film analyses were made. Due to irregularities in the recordings (fig. 30) the transition between stance- and swing-phase was difficult to determine in case 1. In addition, loading conditions were not the same for the two sides. All walking tests in case 1 were performed with simul-

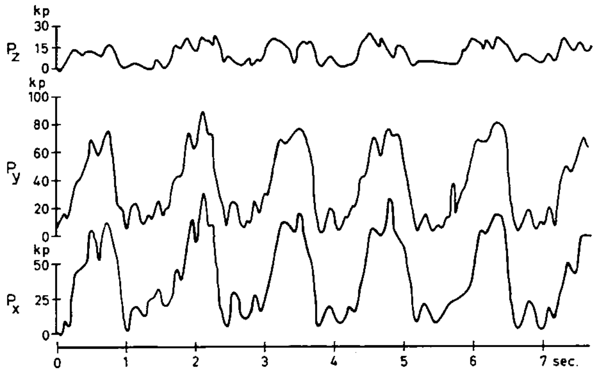

Fig. 30. *The recorded force components from the hip in case 1, level walking. Irregularities occurred in the recordings.*

taneous recordings from the electronic walk-ways. In case 2, all curves were smooth and the transition between the two phases of walking was easily determined from the recordings. When enough information of the difference in loading conditions between the normal and prosthetic side was obtained from the walk-ways, tests were performed on a linoleum-covered floor. Greater walking space was thus available, allowing recordings of several consecutive steps. Determination of the walking speed was more accurate and a faster walking cadence was possible.

In the tests on the electronic floor, the first and the last steps were not included in the data. The recordings on the linoleum were taken for a distance of 7 metres, but the patient started walking several metres in advance of and following the test distance.

Before recording any data the patients accustomed themselves to the walk-ways and to the cable containing the electronic leads, which they carried over their shoulders opposite to the prosthesis.

In the following, the force P and its components will be presented in relation to body-weight W, which was 75 kg in case 1 and 45 kg in case 2.

Case 1

Level walking was performed apparently normally. However, recordings from the electronic floor together with film analyses revealed the differences from normal walking. Though the patient had no pain there was a tendency to unload the prosthetic leg. This unloading was small and the patient was not aware of it. The maximum rise of the vertical force between the foot and the ground in stance-phase was recorded for ten normal persons. In 40 observations the mean difference between left and right side was 0.01 ± 0.06 kp/kg body-weight and $t_{diff} = 0.86$; $p > 0.05$.

In 48 observations on case 1 the mean difference between the normal and the prosthetic side was 0.02 ± 0.09 kp/kg body-weight and $t_{diff} = 1.47$; $p > 0.05$.

When considering the maximum vertical force alone, the unloading of the prosthetic extremity is not great, but other signs of unloading were present. On the diagrams showing vertical force, the time elapsed between heel-strike point *a* and the first maximum *b* and the time between the second maximum *d* and toe-off *e* were absolutely, as well as relatively increased, while the distance between the two maximum force points *b* and *d* was decreased (fig. 31). These changes mean a decreased load per time unit.

In addition, analysis of the film studies shows that the angles between the thigh and the vertical line differ from the normal. As defined on page 71—72 part III the angle *v*, (angle between the vertical line and the thigh of the leg at heel-strike), is usually 25°, and the angle *u*, which occurs simultaneously between the vertical and the opposite thigh, is usually 21°. In case 1, those angles, when the prosthetic leg was at heel-strike, were 23° and 10°, respectively. At heel-strike for the opposite leg, *v* was 24° and *u* 12°.

Recordings were made while walking at two different speeds. The patient was first told to walk at his normal speed. This was found to be about 0.9 m/sec. His fast walking-speed was about 1.3 m/sec. The limited walking space prevented greater speeds.

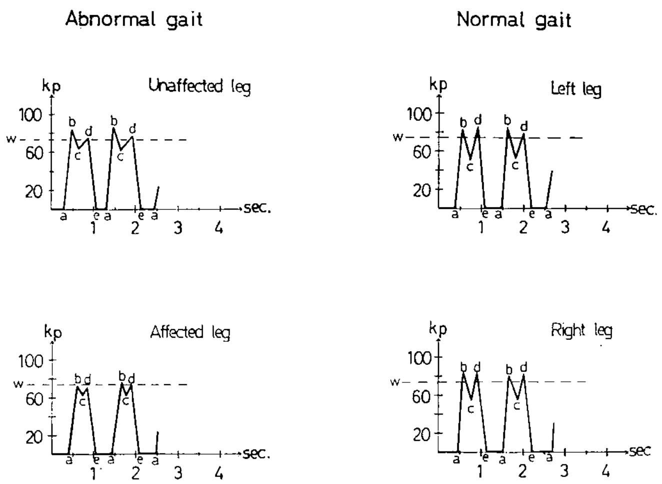

Fig. 31. *The vertical force F recorded by the walk-ways is different in normal and abnormal gait indicating decreased load per time unit on the affected side.*

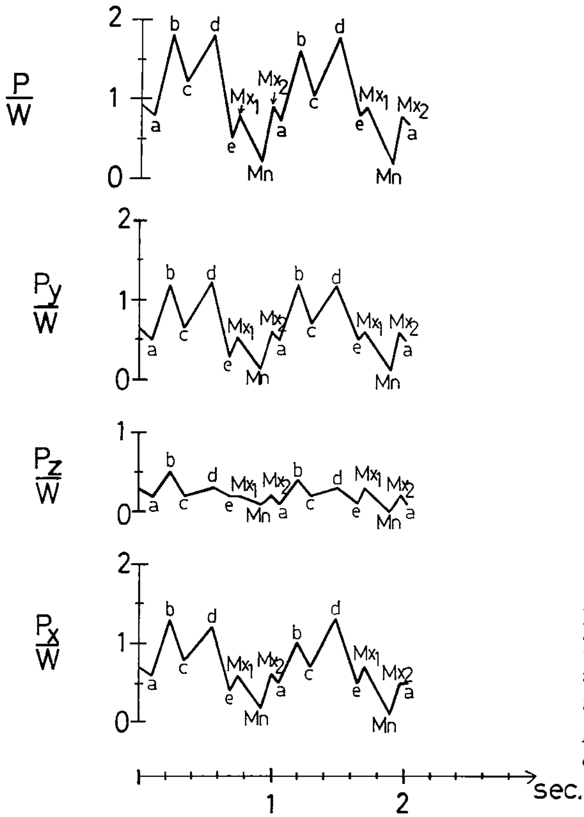

Fig. 32. *Level walking case 1. The force P and its components P_y, P_z and P_x in relation to body weight W. Points refered to in the text are indicated with letters. W=75 kp.*

Recordings from the gauges in section 1 were negligible. In most tests no displacement of the paper travel occurred. The zero-drift of the gauges was of a minor degree and has been considered.

During stance-phase a curve of the three recorded components from the hip was the same shape as the vertical force-curve from the walk-ways (fig. 32). The two maximum points and the low point in between have been designated *b*, *c* and *d*. The points corresponding to heel-strike and toe-off have, as in the vertical force curve, been called *a* and *e*, respectively. Usually they represent low points in the recordings from the hip.

The vertical force curve from the walk-ways and the P_y and P_x curves have points *b*, *c* and *d* coincident in time, while points *b* and *d* on P_z occur 0.01—0.03 sec. later.

The reason for this is not quite clear as the horizontal force occurring between the foot and the ground will have its first maximum point *b* (see fig. 23 page 65) some hundreds of a sec. before *b* but the second maximum point *k* will occur about 5/100 sec. later than *d*. In evaluation of the force P, the small displacement of the points *b* and *d* of P_z have not been taken into account, since they will have no influence of any appreciable size on the calculations.

During the swing-phase forces acted on the prosthetic head. The curves recorded had a characteristic shape fig. 32 revealing two maximum points and a low point between them. The first maximum occurring shortly after point *e* is called Mx1 and the second maximum occurring just before point *a* is called Mx2. The low point is called Mn. The points Mx1 and Mx2 occur simultaneously in all curves, but the point Mn occurs for P_z 0.02—0.05 sec. earlier.

The irregularities occurring in all recordings from the hip-joint are difficult to explain and have not been included in the data. One explanation could be an improper fit of the prosthetic head in the acetabulum. The prosthetic head is 3 mm smaller than the femoral head of the other side. Other possibilities are that the irregularities are due to muscular effects, or to small movements of the stem in the femur. However, the irregularities diminish with increased walking-speed.

Evaluation of the curves was performed in one of two ways. Some were evaluated by use of a vernier scale or by graded transparent plate. Others were done mechanically by automated electronic equipment. Analysis of the curves by the first method included only those points: *a, b, c, d, e,* Mx1, Mn and Mx2. Analysis by the electronic method permitted evaluation of points on the curves separated by an interval of 0.04 sec.

The points *a* and *e*, representing heel-strike and toe-off, are usually low-points but not zero-points in the curves of the hip-joint. Point *e* was negative in only one of thirty-six observations, this was for P_y. Mn was negative five times for P_z, four times for P_y. The negative recordings were small, under 5 kp, except on one occasion when P_y was —9.6 kp.

In table 33 the components and the force P are presented in relation to body-weight at a walking-speed of 0.9 m/sec. and in table 34 the same data are presented at a walking-speed of 1.3 m/sec. The number of observations are 21 and 12, respectively. When the acting forces are great the variation is rather small. The measuring accuracy is low when the recorded figures are small, and this may explain the great variation at those points where the forces are small.

The direction of the forces at the points *b, c, d,* M_{x1} and M_{x2} can be seen in tables 35 and 36. The point of application of the force is suprisingly constant. The head of the prosthesis mainly is subjected to a force coming from above, ventrally and medially.

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
a	0.29±0.10	0.08±0.05	0.32±0.16	0.45±0.16
b	0.97±0.06	0.31±0.06	1.11±0.11	1.51±0.13
c	0.81±0.05	0.15±0.07	0.96±0.11	1.24±0.10
d	1.06±0.09	0.26±0.09	1.15±0.11	1.59±0.11
e	0.12±0.09	0.04±0.04	0.15±0.09	0.20±0.11
Mx1	0.33±0.07	0.10±0.04	0.35±0.11	0.50±0.12
Mn	0.07±0.06	0.01±0.02	0.14±0.09	0.17±0.09
Mx2	0.37±0.10	0.11±0.05	0.43±0.12	0.59±0.13

Table 33. *Case 1. Level walking at 0.9 m/sec. W=75 kp (body-weight). n=21*

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
a	0.55±0.26	0.15±0.09	0.58±0.26	0.84±0.34
b	1.19±0.17	0.48±0.09	1.29±0.22	1.80±0.29
c	0.77±0.11	0.16±0.09	0.82±0.13	1.15±0.14
d	1.20±0.15	0.34±0.06	1.24±0.13	1.76±0.18
e	0.30±0.20	0.15±0.10	0.37±0.15	0.51±0.24
Mx1	0.54±0.22	0.24±0.09	0.59±0.17	0.84±0.28
Mn	0.11±0.10	0.05±0.05	0.17±0.10	0.23±0.12
Mx2	0.57±0.12	0.19±0.05	0.55±0.15	0.85±0.14

Table 34. *Case 1. Level walking at 1.3 m/sec. W=75 kp (body-weight). n=12*

	α°	γ°
b	41±2	18±3
c	42±3	11±5
d	43±3	13±3
Mx1	44±8	17±5
Mx2	42±9	17±6

Table 35. *Case 1. Level walking at 0.9 m/sec. n=21*

	α°	γ°
b	45± 7	22±4
c	44± 9	11±6
d	44± 6	16±3
Mx1	42±10	25±9
Mx2	45±10	19±6

Table 36. *Case 1. Level walking at 1.3 m/sec. n=12*

Because of the low accuracy in recording small forces, the direction of the forces have not been determined at such points. The patient was also tested while walking with a cane. When walking out-doors he usually used a cane as a security against falling. He was told to use the cane in the same way while walking on the walk-ways and recordings were made. When using the cane in the opposite hand there was a considerable unloading of the prosthetic leg. When using the cane in the hand of the prosthetic side, unloading of the opposite extremity was recorded, the patient was not aware of any unloading, regardless on what side he used the cane.

Case 2

Level walking was performed in an apparently normal manner. Recordings from the electronic floor and film analyses revealed only slight differences from the normal. The stance/swing ratio between the opposite and prosthetic side was normal regardless of the walking speed. There were no signs of unloading of the prosthetic leg, but the maximum rise of the vertical force between the foot and the ground was recorded for both sides and in 29 observations the mean difference was 0.02 ± 0.09 kp/kg body weight and $t_{diff} = 1.27$, $p > 0.05$ between the normal and the prosthetic side.
The angles formed by the longitudinal axes of the thighs and the vertical line differed less from the normal than in case 1. In case 2, when the prosthetic leg was at heel-strike, the angle v was 22° and angle u 15° and when the opposite leg was at heel-strike the angles were 22° and 19° respectively. The angle u was slightly less than normal when the prosthetic leg was at heel-strike, (part III, page 72).
Recordings were made at different walking speeds. The most convenient speed was 1.1 m/sec. but when asked she increased the speed up to 1.6 m/sec. Further increase of speed was only obtained while running. The different tests could be grouped in two parts, one group of walking tests at a speed of about 1.1 m/sec, and another group at about 1.4 m/sec. The results are therefore presented in mean values and standard deviations from these two groups.
Recordings from the gauges in section 1 have always been taken into account and all readings were made optically. No zero-drift occurred. The shape of the curves (fig. 33) differed somewhat from those of case 1. During the stance-phase two maximum points and a low point in between occurred in accordance with the curve F and recordings from case 1. The points corresponding to Mx_2 in the swingphase and to point a at heel-strike were represented by a plateau of the curves. The points

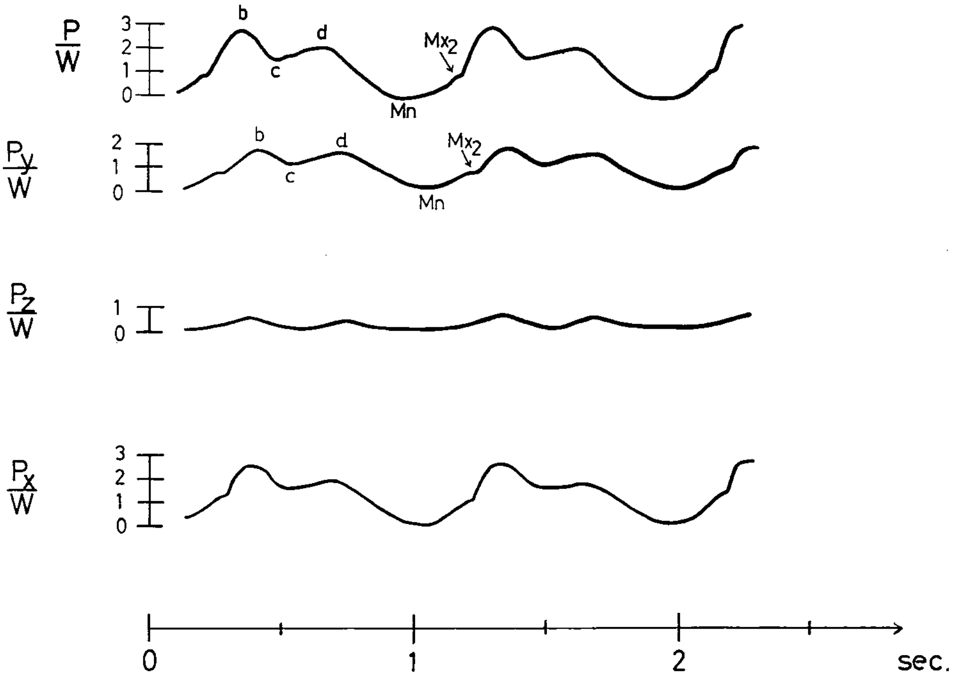

Fig. 33. *Level walking case 2. The force P and the components P_y, P_z and P_x in relation to body weight W. Points refered to in the text are indicated with letters. W=45 kp.*

corresponding to toe-off and Mx1 were usually not represented by any changes in the curves. Sometimes a decrease of the slope of the curves was recorded.

In case 2 the stance-phase was characterized only by the points *a*, *b*, *c* and *d*. The point *a*, however, was not a low point but identical to point Mx2 in the swing-phase. The points *e* could sometimes be seen as a decreasing slope of the curve, but usually the curve continued to fall smoothly from point *d* to the low point of the swing-phase, Mn. At this point the force curves were always close to zero. Sometimes a negative force of 1 or 2 kp was recorded from the curves P_z and P_x. This magnitude was of negligible size. At the end of the swing-phase, point Mx2 occurred as a horizontal run of all the curves, perhaps followed by a low point representing point *a* which was of non-measurable magnitude. The same changes of the curves at points *a*, *e*, Mx1 and Mx2 recorded in case 1 occurred while walking with the knee stiff and while running (page 103).

The vertical force curve F from the walk-ways and the P_y and P_x curves have points *b*, *c* and *d* coincident in time, while points *b* and *d* on P_z, as in case 1 occurred 0.01—0.03 sec. later.

In table 37 the components and the force P are presented in relation to body-weight at a walking speed of 1.1 m/sec, and in table 38 the same data are presented at a walking speed of 1.4 m/sec. The number of observations are 24 and 18 respectively. The direction of the forces at points *b*, *c*, *d* and Mx2 is seen in tables 39 and 40.

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
b	1.52±0.08	0.50±0.04	2.49±0.15	2.95±0.16
c	0.94±0.21	0.06±0.04	1.54±0.11	1.86±0.12
d	1.35±0.08	0.26±0.06	1.74±0.13	2.23±0.12
Mx2	0.64±0.09	0.12±0.03	1.02±0.12	1.21±0.13

Table 37. *Case 2. Level walking at 1.1 m/sec. W=45 kp (body-weight). n=24*

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
b	1.71±0.17	0.52±0.09	2.74±0.28	3.27±0.32
c	1.09±0.10	0.04±0.06	1.60±0.20	1.94±0.21
d	1.56±0.12	0.32±0.06	1.98±0.17	2.55±0.19
Mx2	0.72±0.12	0.10±0.03	0.88±0.15	1.15±0.18

Table 38. *Case 2. Level walking at 1.4 m/sec. W=45 kp (body-weight). n=18*

	α°	γ°
b	31±1	18±1
c	34±2	3±3
d	38±2	11±3
Mx2	34±4	10±2

Table 39. *Case 2. Level walking at 1.1 m/sec. n=24*

	α°	γ°
b	32±1	17±3
c	34±2	2±3
d	38±2	12±2
Mx2	39±3	8±3

Table 40. *Case 2. Level walking at 1.4 m/sec. n=12*

The point of application of the force is rather constant and differs little from case 1. The head of the prosthesis is subjected to a force directed against a rather small area of the medial, upper ventral surface.

Stair-walking

The force acting on the prosthetic head while walking up- and downstairs was determined. The stairs were made of stone and each step was 15 cm high. In case 1 the stairs were 270 cm long and in case 2 the stairs were 360 cm long. In the analyses, recordings from the first step and the last step have not been included.
Before recording any data the patients climbed the stairs several times to become accustomed to them. During the tests, the cable containing the leads was carried over the shoulder opposite to the prosthesis.
No recordings with force-plates were made. This meant that the transition between the stance- and swing-phase could not be determined.
In the following, the force P and its components are presented in relation to the bodyweight W. In case 1, W equals 75 kg and in case 2, 45 kg.

Walking upstairs. Case 1.
In case 1, stair-walking was performed with ease. Because he always limped on the first two steps, the recordings started two steps below the measured distance. In all tests the walking-speed was 0.6 m/sec. If he altered this speed, the gait was slightly affected.
Recordings from section 1 were negative. In the readings consideration had to be taken of a minor zero-drift and displacement in the travel of the recording paper. The zero-drift was most obvious regarding P_y. The zero-line was therefore determined individually for each step.
The shape of the recorded curves in the stance-phase was similar to that for level walking. Though the exact transition between the stance- and swing-phase could not be determined, the maximum and minimum forces of the two phases could be separated.
As in level walking, two maximum points, *b* and *d* and a minimum point *c* which occurred between *b* and *d*, were present in the stance-phase (fig. 34). In walking upstairs, however, the *b- c- d*-part of the curve was narrower and the force representing the point *b* was always considerably greater than at point *d*. This was most pronounced for the component P_x.
Compared to level walking, the swing-phase revealed a different curve. In walking upstairs two minimum points, Mn_1 and Mn_2, and a single maximum point Mx occurred during the swing-phase (fig. 34). Points Mn_1, Mn_2 and Mx were coincident on the time axis for P_y and P_x. For P_z, Mn_1

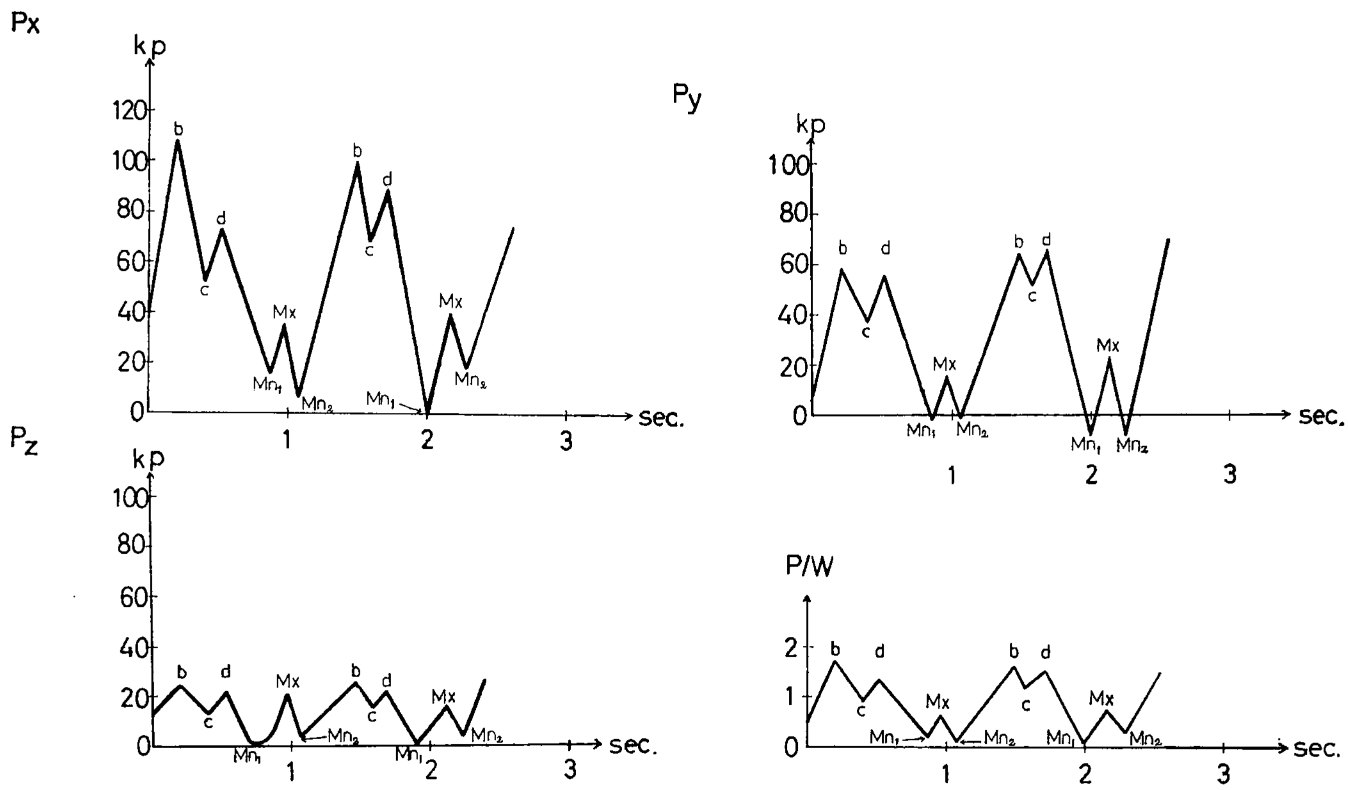

Fig. 34. *Walking upstairs case 1. The components are presented in kp and the force P in relation to body weight W. Points refered to in the text are indicated with letters. W=75 kp.*

occurred about 1/10 sec. earlier. The forces approached zero at the points Mn_1 and Mn_2 and P_y sometimes was negative. As mentioned previously the accuracy of measuring small forces was low and the results obtained under these conditions must be weighed with caution.
In table 41 the recorded forces are presented. The variation is great when the forces are small. The variation of the direction of forces is small when the force is great. The direction of the forces corresponding to the points *b, c, d* and Mx can be seen in table 42. The direction of force at points Mn_1 and Mn_2 varied widely and values were considered to be unreliable. It should be emphasized that at the points Mn_1 and Mn_2, the force P_y was small (under 5 kp) and sometimes negative. The force P_z was always positive and varied between 3—15 kp. P_x was consistently positive and varied between 0—17 kp. The head of the prosthesis during the swing-phase was occasionally subjected to a small force coming from below and ventrally, instead of the usual direction.
In one of the tests the patient slipped and nearly fell. The recorded force was about 180 kp and α 38° and γ 26°. In the following swing-phase there was a negative value for P_x of about 16 kp.

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
b	0.86±0.06	0.33±0.06	1.27±0.17	1.54±0.13
c	0.64±0.08	0.18±0.08	0.79±0.15	1.04±0.16
d	0.80±0.07	0.23±0.08	0.99±0.13	1.30±0.14
Mn_1	—0.01±0.05	0.11±0.08	0.14±0.08	0.21±0.05
Mx	0.18±0.10	0.23±0.06	0.38±0.14	0.49±0.13
Mn_2	—0.02±0.07	0.09±0.04	0.09±0.10	0.17±0.07

Table 41. *Case 1. Walking upstairs. W=body weight 75 kp. n=9*

	$\alpha°$	$\gamma°$
b	35± 5	21± 4
c	39± 5	15± 6
d	40± 3	16± 5
Mx	30±11	50±12

Table 42. *Case 1. Walking upstairs. n=9*

Walking upstairs. Case 2.
The patient walked upstairs apparently normally and was able to vary her walking-speed. When the patient was asked to climb the stairs slowly, she did so at a speed of 0.6—0.8 m/sec. and the curves had the same shape

as in case 1. When the patient was told to walk faster her walking-speed was 0.9 m/sec. but during the stance-phase the curves developed a different shape. As in running the points *b*, *c* and *d* were replaced by a single peak, *bd* (fig. 35).

The readings were easily recorded as no zero-drift or displacement of the recording paper occurred. As in case 1, the exact transition between the stance-phase and the swing-phase could not be determined. In the tests when the walking-speed was slow and both points *b* and *d* were distinct the force acting at point *b* was considerably greater than at the point *d*. The force acting at point *b* was of about the same magnitude as the force acting at point *bd* at the higher walking-speed, and therefore point b has been analysed together with point *bd*.

In the swing-phase the same displacement of point Mn_1 for the force P_z as in case 1 occurred. P_z showed at Mn_2 a steeper slope and P_z was sometimes negative. P_y and P_z were maximum at point Mx but for P_x it was only a point of inflection or had a less steep slope. As P_x was the greatest component of the force P, it did not always have a maximum at point Mx. Table 43 shows the forces acting on the prosthetic head and table 44 shows the direction of the forces at the points *bd* and Mx. As in case 1, the direction of the force at the points Mn_1 and Mn_2 cannot be determined with accuracy. No negative values were recorded for P_y, but P_z was negative in six of fourteen steps at point Mn_2. The negative forces were small and did not exceed 3 kp. The head of the prosthesis was thus occasionally subjected to a small force momentarily applied, coming dorsally from above.

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
bd	1.72±0.16	0.44±0.04	2.87±0.19	3.38±0.18
Mn_1	0.23±0.07	0.20±0.05	0.36±0.12	0.48±0.12
Mx	0.26±0.07	0.20±0.05	0.35±0.12	0.46±0.16
Mn_2	0.12±0.06	0.01±0.05	0.10±0.11	0.20±0.09

Table 43. *Case 2. Walking upstairs. W=body-weight 45 kp. n=14*

	$\alpha°$	$\gamma°$
bd	31±3	15±2
Mx	37±9	37±5

Table 44. *Case 2. Walking upstairs. n=14*

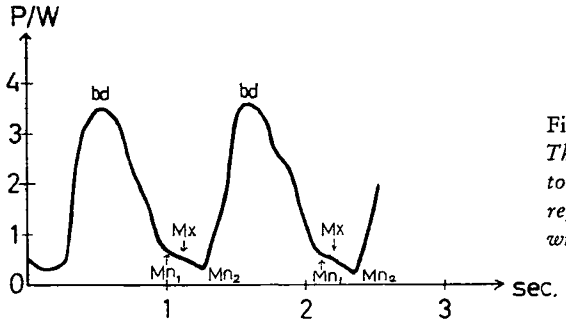

Fig. 35. *Walking upstairs case 2. The force P is presented in relation to body weight W and the points refered to in the text are indicated with letters. W=45 kp.*

Walking downstairs. Case 1.
The patient walked downstairs apparently normally, but subjectively he felt insecure compared to walking upstairs. The walking-speed was constant in all tests and any attempt to alter it affected the gait.
The recordings from section 1 were negligible. A slight zero-drift and displacement in the travel of the recording paper occurred as while walking upstairs and has been taken into account. The shape of the curves were the same walking downstairs as in walking upstairs both in the swing-phase and the stance-phase. In the stance-phase, however, there was a difference in magnitude as the force at point *d* always was greater than at point *b* (fig. 36). The point Mn_1 for the force P_z appeared 1/10 sec. earlier than for P_y and P_x. At the points Mn_1 and Mn_2 the force P_y was always negative, and small. The point Mx was always maximum for all components.
In table 45 the forces acting on the prosthetic head are presented. In table 46 the direction of the forces at the points *b,c,d* and Mx are shown. Because of the low degree of accuracy in measuring small forces, the directions of forces at the points Mn_1 and Mn_2 are not presented. When the force P_y is negative at these points, the head is subjected to a force coming from below and ventrally.

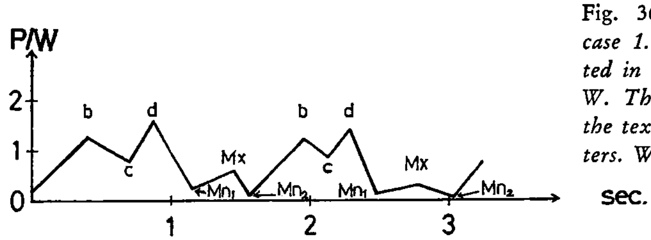

Fig. 36. *Walking downstairs case 1. The force P is presented in relation to body weight W. The points refered to in the text are indicated with letters. W=75 kp.*

	P_y/W	P_z/W	P_x/W	P/W
b	0.80±0.05	0.27±0.07	0.96±0.09	1.29±0.11
c	0.60±0.11	0.17±0.08	0.71±0.18	0.95±0.19
d	0.95±0.09	0.39±0.09	1.21±0.18	1.59±0.18
Mn1	—0.12±0.05	0.12±0.04	0.08±0.07	0.20±0.02
Mx	0.07±0.06	0.22±0.09	0.32±0.12	0.40±0.13
Mn2	—0.04±0.02	0.11±0.05	0.12±0.08	0.17±0.08

Table 45. *Case 1. Walking downstairs. W=body-weight 75 kp. n=8*

	α°	γ°
b	40±2	19±5
c	40±6	16±6
d	38±3	23±5
Mx	17±9	68±8

Table 46. *Case 1. Walking downstairs. In one recording at point Mx the angle* α *was —22° and angle* γ *was —50°. These values are not included. n=8*

Walking downstairs. Case 2.

The patient walked apparently normally and was able to vary her walking-speed. The shape of the curves was the same as in walking downstairs, case 1. The force acting at point *d* was considerably greater than at point *b* (fig. 37).

The readings from section 1 have been taken into account in the determination of the forces. No zero-drift or displacement of the recording paper occurred and no negative values were recorded.

In table 47 the recorded forces are shown and in table 48 the direction of the forces at points *b, c, d* and Mx are shown. In all tests the head of the prosthesis was subjected to a force coming from above, ventrally and medially.

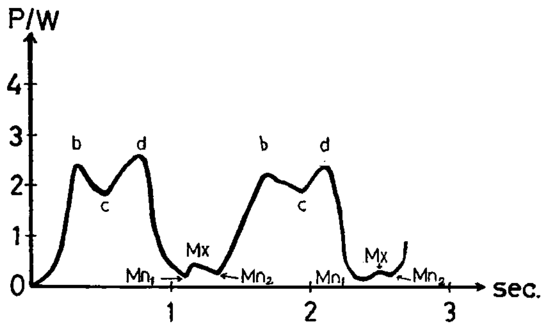

Fig. 37. *Walking downstairs case 2. The force P is presented in relation to body weight W. The points refered to in the text are indicated with letters. W=45 kp.*

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
b	1.59±0.25	0.47±0.10	2.21±0.35	2.77±0.41
c	1.32±0.15	0.33±0.06	1.79±0.33	2.25±0.59
d	1.75±0.11	0.43±0.07	2.17±0.34	2.83±0.31
Mn_1	0.12±0.08	0.04±0.03	0.11±0.07	0.18±0.06
Mx	0.29±0.16	0.08±0.04	0.26±0.14	0.42±0.17
Mn_2	0.21±0.16	0.05±0.04	0.17±0.13	0.30±0.16

Table 47. *Case 2. Walking downstairs. W=body-weight 45 kp. n=12*

	α°	γ°
b	36± 3	17± 3
c	37± 4	14± 2
d	39± 4	14± 2
Mx	47±20	21±17

Table 48. *Case 2. Walking downstairs. n=12*

Discussion

While walking up- and downstairs, the curves of the force components in the stance-phase have a shape similar to those in level walking and running. In the swing-phase, however, the shape of the curve differed, in that it had low points with a maximum in between.

While walking upstairs the force acting at point *b* was considerably greater than at point *d* and in case 2 when the speed was increased the points *b*, *c* and *d* fused to one point *bd* as occurred while running.

While walking downstairs the point *d* in the stance-phase was considerably greater than point *b*.

In all tests the point Mn_1 in the swing-phase appeared 1/10 sec. earlier for P_z than for the other two components. In case 1 the force P_y frequently was negative which could be due to errors in the readings. When negative, the forces were always small. In case 2, while walking upstairs, P_z was negative occasionally at point Mn_2. Point Mx was always a maximum point for the components P_y and P_z but in walking upstairs for P_x it is a point where the curve changes its slope. Because P_x usually is the main component of the force P, Mx does not always correspond to a maximum point on the force diagram of P.

When the force was great, as at points *b*, *c* and *d*, and sometimes at point Mx, the direction of force varied slightly and the prosthetic head was loaded in the usual area (superiorly, ventrally and medially). At point

Mx the direction of the force becomes more horizontal and has greater variation.
The shape of the curves and the direction of the forces were in agreement in the two cases. The magnitude of the forces was however different. The force on the prosthetic head relative to body-weight was greater in the stance-phase of case 2. Theoretical calculations indicate a higher load to occur in the stance-phase of case 2. The test results show the load difference to be greater than those calculations indicate.

Running

Running causes the force acting on the femoral head to be of high magnitude. The following information is on measurements performed while running.
Case 1 was afraid to run, therefore no tests were performed.
Case 2 was able to run easily. The patient stated she had not run for the last 30 years, but was able to do so without difficulty. When she became accustomed to running the recordings were started.
The running tests were first performed on the walk-ways to determine the transition between the stance- and swing-phase. Then the tests were repeated on a linoleum-covered floor of greater space. The recordings on the linoleum were taken for a distance of 7 metres, but the patient started to run several metres in advance of and continued to run after completing the test distance.
Running was performed at a speed of about 2.0 m/sec, and could be easily increased to 3.0 m/sec. Increasing the speed beyond this was deemed inadvisable.
No significant differences in the magnitude of forces were obtained at different speeds. Analyses have been made for 19 steps. The original curves have been analyzed in a computer, which gives the forces and angles numerically and graphically (fig. 38).
Evaluation of the curves has considered the recordings of section 1.
The vertical force F, recorded by the walk-ways measuring devices has only one peak in running here called *bd*. The forces recorded from the hip device has the same peak *bd* in the stance-phase. This peak occurred simultaneously for F, P_x and P_y. For P_z, however, the point *bd* occurred 0.02—0.04 sec. later. All readings for point *bd* were made when P_x and P_y reached this peak, which means that the values for P_z were taken just before its peak. The difference thus occurring is negligible except for the angle γ, which at point *bd* of P_z would be significantly higher.

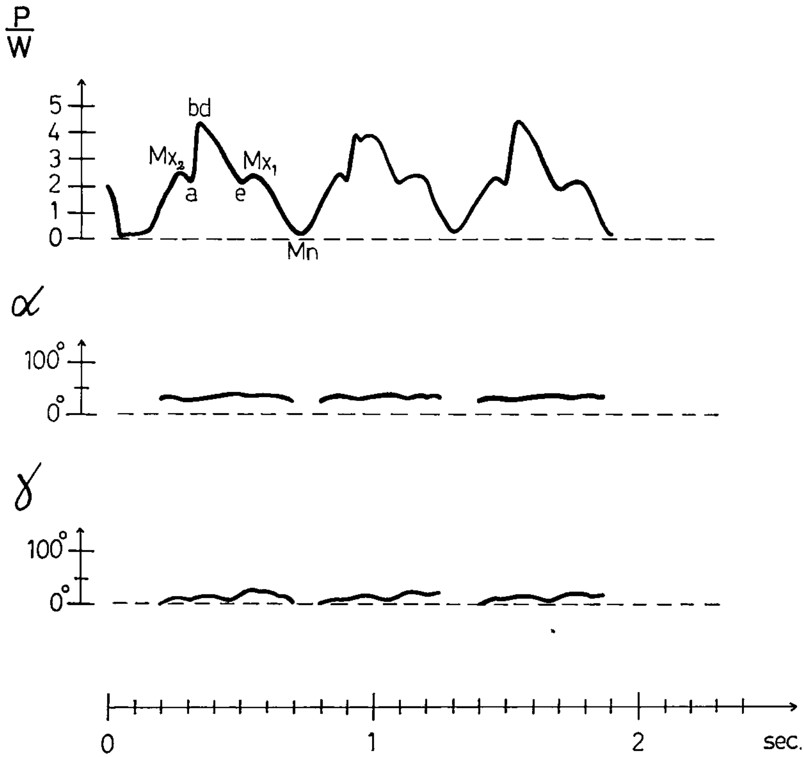

Fig. 38. *Running case 2. The force P is presented in relation to body weight W. The variation of the angles* α *and* γ *in the stance-phase is indicated. W=45 kp.*

As in level walking two maximum points Mx_1 and Mx_2 and a low point between the two peaks occurred in the swing-phase. Mx_1 and Mx_2 were however of much higher magnitude than in level walking. At point Mn the readings for P_z were always negative and for P_x and P_y positive. Recorded values at the points *a* and *e* were also of high magnitude. In Tables 49 and 50 forces and angles obtained are presented. At point Mn the direction of the force is not certain because the acting forces are small and the accuracy of recording low. The mean value of α was 15° and γ —66° at Mn, but varied greatly.

The angle γ is small, about 10° at point Mx_2 and *a*, and then increased throughout the stance-phase to reach its greatest value 25° at Mx_1. This fact is contrary to that which was expected.

	$\frac{P_y}{W}$	$\frac{P_z}{W}$	$\frac{P_x}{W}$	$\frac{P}{W}$
a	1.21±0.13	0.21±0.03	1.92±0.25	2.28±0.27
bd	2.26±0.17	0.59±0.09	3.65±0.29	4.33±0.32
e	1.28±0.22	0.42±0.08	1.81±0.34	2.26±0.39
Mx1	1.28±0.15	0.60±0.12	2.01±0.39	2.49±0.44
Mn	0.04±0.04	—0.09±0.04	0.15±0.11	0.20±0.08
Mx2	1.30±0.11	0.23±0.03	2.09±0.20	2.47±0.22

Table 49. *Case 2. Running at about 2.5 m/sec. W=45 kp. n=19*

	$\alpha°$	$\gamma°$
a	32±3	10±1
bd	32±2	15±2
e	35±4	19±5
Mx1	33±4	25±3
Mx2	32±2	10±1

Table 50. *Case 2. Running at about 2.5 m/sec. W=45 kp. n=19*

Coefficient of friction

The measuring prosthesis enables us to get an approximate value of the coefficient of friction. The theories regarding this are presented in part II page 54—57. Determination of this quantity is dependent on the size of the contact area and it will be accurate only if the contact area is not too great. If the contact area is large the recorded values will be too high. The diameter of the femoral head is normally less than that of acetabulum and in case 1, this differential is greater since the prosthetic head is about 3 mm smaller than the original head. In case 2 the diameter of the prosthetic head is about the same as that of the original head. The contact area is therefore probably greater in case 2.

Case 1

As described in part III page 57 the torque M_x was not recorded. This introduces an error but the error does not exceed 20 %. If M_x is smaller than M_{1y} and M_{1z}, as it usually was in case 2, the error is less.

The coefficient of friction was determined under dynamic conditions while walking and under static conditions while standing on one leg and keeping the leg flexed against a resistance.

While walking, the coefficient of friction changed after about 25 metres. The patient at the same time complained of fatigue but not pain in the hip. The coefficient of friction returned to its usual value after resting. The dynamic coefficient of friction was found to be 0.021 ± 0.001 in 12 tests and after walking about 25 metres 0.033 ± 0.003.
The static coefficient of friction was markedly increased. Standing on one leg gave 0.115 ± 0.016 after 30 readings and in the supine position when flexing and abducting the leg the coefficient was 0.102 ± 0.018.
The tests performed under static conditions may not always produce maximum friction so that the real values would be somewhat higher than the mean value.

Case 2

In this case M_x was recorded as well as M_{1y} and M_{1z}.
The coefficient of friction could not be determined under static conditions as no movements occurred. From the gauges of section 1, only a single steady deflection occurred under load, below the frictional forces.
While walking a maximal moment about the y-axis developed in the middle of the stance-phase and ended in the beginning of the swing-phase. This moment is probably generated by catching of the prosthetic head against the ventral ligaments of the joint and is not due to friction. The coefficient of friction in this case was determined from point *b* in the stance-phase and from Mx_2 in the swing-phase.
The value obtained in 36 tests was 0.042 ± 0.013. The inherent error in determining the coefficient of friction is greater in case 2 than in case 1 since the head-acetabulum diameter ratio in case 2 gives a greater contact surface.

V. Recapitulation and discussion

Analyses of forces acting on bone, stress and strain in bone under load and determination of physical properties of bone have been made by several anatomists and orthopaedic surgeons. However, theoretical calculations are complicated as bone is a heterogenous material and of complex geometric shape. Much of the force is due to muscular activity and it is difficult to determine quantitatively. Analyses, even with small demands upon accuracy, can only be done under idealized conditions.

Actual measurements of forces with their stresses and strains will provide more accurate results and enable theory to be confirmed or altered. Specimen measurements have informed us of stress-concentrations and fracture mechanisms. Specimen testing is limited since the force vectors acting under vital conditions are unknown. The only possible way to get information regarding the magnitude and direction of physiological forces is to record them under vital conditions.

At present it is impossible to record forces under pure physiological conditions because introduction of the measuring device will disturb the normal state. This must be taken into consideration when evaluating the results of such recordings.

The measuring prosthesis

The purpose of this work was to measure forces acting on a prosthetic head of femur in the active patient. At the start it was clear that quantitative measurements on living bone would be impossible, with the techniques available today. This has been done on specimens (Hirsch and Frankel 1960, 1961). The proximal part of the femur is quite often replaced by a metal prosthesis and since metal is a highly suitable material for support of a strain gauge, it was thought that placing strain gauges into a prosthetic neck would provide an easy way to measure forces acting on the hip. Strain gauge placement on the neck of a prosthesis permits that prosthesis to act as a force transducer. Three orthogonal force components are recorded from this device which permits the resultant force to be computed. A femoral endo-prosthesis, acting as a three-dimensional force transducer while maintaining its clinical purpose was proposed.

From an electronic viewpoint the construction of a measuring prosthesis was not difficult. The neckpart became the transducer and the strain gauges were applied as in a six component strain gauge balance.

Strain gauges are sensitive to moisture and temperature changes so they were placed inside the neck where they were protected from body fluids and where the temperature would remain constant.
Problems which arose were purely of technical origin. The neck of the prosthesis had to be small and the area available for the strain gauges was limited. If the measurements were to be accurate the dimensions of the prosthesis had to be exact. The material requirements for the prosthesis were easy machinability with adequate strength to permit small dimensions. The prosthetic shape had to fit anatomical structures and the material be without toxic effect.
Design of the prosthesis had to account for the manner in which the signals could be transmitted to the recording unit.
The connections between the neck, head and stem had to be completely waterproof.
The metals usually used for surgical implants were rejected, because they were too weak (18/8 Mo steel), too difficult to machine (cobolt-chromium alloys) or had low wear resistance (titanium). If a small amount of titanium is added to stainless steel its strength is increased as well as its resistance to corrosion. A titanium stabilized stainless steel was used for the measuring prosthesis.
The shape of the prosthesis had to be adapted to anatomical requirements. It could not be made especially for the person who would use it, therefore the shape and outer dimensions were determined from mean values. The diameter of the prosthetic head should be the same as the femoral head it replaces. The size most often used in our experience is $1^7/_8$ inch or 47.2 mm. The diameter of the head of the measuring prosthesis was chosen to be 47.2 mm.
The length of the neck should be as long as possible to deliver strong signals. Both the Moore and Thompson prostheses have a distance of 48—50 mm between the vertex of the sphere and the inferior surface of the supporting plate against the neckrest, when the head diameter is 47.2 mm.
Wertheimer and Martin (1963) determined the distance between the vertex of the femoral head and the point of the outer surface of the femur cut by the longitudinal axis of the neck in 80 specimens. They found extreme measurements to be 75 and 110 mm but no information regarding the variation was given. No appreciable difference between left and right side of the same subject was found. The mean between the two extremes is 93 mm. Taking these figures and manufacturing factors into consideration, the distance between the vertex of the measuring prosthesis and the supporting surface against the neck rest was chosen to be 68.6 mm.

If the distance between the neck rest and the lateral surface of the femur along the longitudinal axis can be approximated to about 25 mm, a total length of 93.6 mm occurs.
The length of the neck was 30 mm, but if the head was made hollow, the neck could pass 10 mm beyond the centre. Gauges could be placed in a section through the midpoint of the sphere.
The angle between the stem and the neck part of the prosthesis was chosen to be 120°. The true cervico-diaphyseal angle is a few degrees greater. In both cases, however, the cervico-diaphyseal angle was found to be smaller than that of the opposite side. From an anatomic as well as from a recording point of view the cervico-diaphyseal angle of the prosthesis should have been greater.
Transmission of the signals from the prosthesis to the recording unit was discussed. From the beginning it was thought that telemetry would have been the best method, but it had to be rejected for economical reasons. It was decided to have the signals transmitted by leads. The leads contained toxic material so they were covered with terylene and teflon. At the distal end the leads were connected to a hermetically sealed contact house.
The recordings could not begin for a period of time after surgery so the leads and the contact house were left under the fascia for about 6 months. Then the contact house was brought out through an incision.
When the recordings were completed the leads were removed, close to the prosthesis because they were potentially toxic. The prosthesis was provided with a sharp edge at the junction of the leads and the prosthesis, in order to cut them free by pulling them against this edge. Removal by this method failed, because the angle between the leads and the edge was not great enough. Removal of the leads required a separate operation.
Two measuring prostheses have been manufactured and placed into two patients. Recordings were begun about six months after the operation. The recordings were completed within a week because of the risk of hip-joint infection secondary to the lead tract from the skin surface.

Electronic walk-ways

The use of a femoral end prosthesis creates abnormal conditions, which might disturb the function of the hip-joint. From experience it is known that the functional results in patients with such prostheses varies. There will be patients who have an apparently normal gait and some with a marked limp. An apparently normal gait does not necessarily mean that the gait is normal. A method which makes it possible to determine the load bearing capacity of the lower extremities has been developed. The walk-ways and filmtrack described in Part III were constructed to determine the transition between stance- and swing-phase in both walking and running. This is important in the analysis of the forces which act on the hip-joint.

By means of force transducers the walk-ways record the vertical and the horizontal forces (Part III, page 65). From the vertical force curves, differences of load intensity between the left and the right side could be determined as well as the transition between the stance- and the swing-phase.

The walk-ways have a relatively low natural frequency of about 50 cps. Force-sequences of frequency greater than $^1/_8$ of the natural frequency or about 6 cps are not recorded with high accuracy. With the exception of the force g of the horizontal curves, (Part III, page 65) force-sequences of more than 6 cps were not recorded. Influence of the recordings by the natural frequency of the walk-ways is of minor importance since comparisons between left and right side only have been made.

Recordings from persons without gait disturbances revealed no difference between the left and right sides. Gait disturbance changes the shape and magnitude of the vertical force curves, so that in limping, the vertical force is usally smaller on the affected side. Sometimes the maximum value of the vertical force is equal on both sides, even though both extremities do not carry equal loads. The area under the vertical force curve in the stance-phase is always diminished in limping, which is a sign of less load per time unit.

A limping person always remains longer on the non-affected side and his stance-phase is diminished on the affected side and prolonged on the good side. This means that the swing-phase must be prolonged on the affected side and diminished on the good side. The created difference between the ratio of stance-phase to swing-phase between the two sides becomes a sensitive sign of gait disturbance. Another alteration is observed when the distance between the two maximum points *b* and *d* (part III, page 89 fig. 31) is diminished. The foregoing changes are recorded by means of the walk-ways.

The ratio between the stance-phase and the swing-phase varies with different walking-speeds.

Normally no difference in this ratio between the left and the right side should occur and this index should be 1.0. Tests on normal persons confirmed this. Persons with gait disturbances have an index greater than 1.0, especially during slow-walking when values exceed 1.5. As the walking-speed increases, the index decreases, and if the gait is fast enough the index approaches 1.0. The decreased distance between the points *b* and *d* is not influenced by the walking-speed.

The two persons equipped with a measuring prosthesis, case 1 and case 2, have been tested on the walk-ways. Differences from normal were found to exist in both cases but they were more pronounced in case 1 where a slight difference of a maximum load of the force F was recorded (Part III page 89). The area this force-curve formed in the stance-phase was decreased. The index of the ratio stance-phase to swing-phase between the opposite and the prosthetic sides was increased, but during rapid walking the index approached 1.0. The diminished distance between the point *b* and *d* persisted in all tests.

In case 2 there was small difference of the maximum value for the vertical force F, but the index of the ratio stance-phase to swing-phase and the distance between the points *b* and *d* were normal.

Simultaneous recordings of the vertical force F from the walk-ways and the forces recorded from the hip-joint, showed a correlation in time between their maximum and minimum values for the stance-phase. The transition between the stance- and swing-phases occurred with changes in the shape of the curves recorded from the prostheses. This was more pronounced during running.

Recordings from the walk-ways were correlated to cinema-photography by means of a synchronized watch so that each film frame could be matched to a given point on the force curves. The film was exposed at 64 frames per second.

When the direction of the forces acting on the hip-joint are considered it must be kept in mind that all forces are measured in relation to a coordinate system relative to the prosthesis. This makes it necessary to know the size of the angle formed by the thigh and the perpendicular line at heel-strike and toe-off. This angle might differ from the normal in cases where a femoral end prosthesis is present.

The angle (part III, page 71, fig. 27) which the longitudinal axis of the thigh forms with the perpendicular line, the angle v, and the angle occurring at the same time between the dorsal thigh and the perpendicular line, the angle u, have been determined. During normal gait the angle v is about 24° and the angle u about 21°.

In case 1, the angle v was 24° when the non-operated leg was at heel-strike and in case 2 it was 22°. These values are normal. The angle u at the same time on the prosthetic side was 12° and 19°, respectively. The value for angle u obtained in case 1 is not normal. When the prosthetic leg was at heel-strike the angle v in case 1 was 23° and in case 2 it was 22° and the angle u at the same time on the non-operated leg was 12° and 15°, respectively. The values obtained for angle u are not normal for either case 1 or case 2.

Evaluation of the recordings

In evaluating the recordings, the new anatomical conditions due to the prosthesis and reading errors of the curves must be taken into consideration.

The only way to get information about the original cervico-diaphyseal angle on the operated side is to measure this angle on the opposite side. The angles are not necessarily the same for the left and the right sides. Wertheimer and Martin (1963), after studying 80 femurs, found the extreme variation between the left and the right side to be 11°.

In case 1 the cervico-diaphyseal angle of the non-operated side was 130°, and in case 2, 140°.

Normally there is an antetorsion of the upper femur. The antetorsion angle of the non-operated and the prosthetic sides have been determined. Wertheimer and Martin (1963) found the extreme variation between the two sides in the same subject to be 29°. On the non-operated side the antetorsion angle was found to be 0° and 38° in cases 1 and 2, and on the prosthetic side, 8° and 35°, respectively. This angle showed good agreement between the prosthetic and opposite sides.

The differences from the normal recorded by the walk-ways and the film-equipment must be taken into consideration when the results are evaluated. The force components were recorded as curves on a graph and the magnitude at different points of the curves was measured from a zero-line. In this work the readings have been done in several ways. The initial measurements were made by use of a vernier and the procedure was very time consuming. A transparent plate graduated in mm was more convenient. Some dynamic sequences which demanded readings with short intervals were analyzed in an optical reading machine which collects the results on a perforated tape. The tapes were then analyzed in a computer which described the forces numerically and produced new curves to a common scale. Errors are inherent in all the methods of reading. Control-tests have shown it possible to read the curves with an accuracy of $\pm$ 0.2 mm. When small deflections occur the error is somewhat increased.

Since the force corresponds to a linear measure, the reading error depends upon the ratio between the length unit and force unit. The error varies for different curves, and for each curve the error increases as the force diminishes. For P_y and for P_z a reading of 0.2 mm represents a force of about 2 kp, but for P_x it represents a force of about 5 kp. As the readings are less precise in small deflections, quantitative determination of forces under 10 kp for P_y and P_z and under 25 kp for P_x are of low accuracy. As can be seen from the result, the standard deviations are greatest for small forces. In calibration tests of the prosthesis, a digital voltmeter showed an error of less than 5 % in the determination of forces and angles. The galvanometers used respond to frequences of up to 60 cps with an accuracy of $\pm$ 5 % and low frequencies are recorded more accurately.

Intravital measurements and results

In this study, forces acting on the head of the measuring prosthesis have been determined as to magnitude and direction. The results are valid for the prosthetic hip and to apply these results in a general way to normal hips the application must be made with caution and conditions secondary to implantation of the prosthesis must be regarded.

From an anatomical viewpoint the new angles and dimensions must be compared to those existing before the operation. Since these factors are unknown, those of the opposite hip have been used for comparative purposes.

It is very difficult to precisely determine the lever arms and the direction of the muscle pull. One of the few situations where theoretical calculations can be performed with a high degree of accuracy is for one-leg support, when the force is obtained in only one plane.

Theoretical calculations for one-leg support provide the following information. In case 1, the force calculated was 1.8 times the body-weight, when standing on the prosthetic leg and 2.0 times when standing on the opposite leg. The same forces for case 2 were 2.0 and 2.3 times the body weight respectively. The actual force recorded in case 1 from the prosthesis, while standing on the prosthetic leg, was 2.3 times the body weight and, in case 2, it was 2.8 times the body weight. The reason the recorded values were higher than the theoretical may be due to muscular forces. The actual angle formed by the force with the longitudinal axis of the neck was less than that calculated and this may also be due to muscular effects.

The proximal femur has a longitudinal axis which does not coincide with the ideal longitudinal axis of the femur. The angle (part I, page 10) formed between these two axes opens ventrally and according to Backman it is $8.0^\circ \pm 1.9^\circ$. If the direction of the recorded forces are taken relative to the longitudinal axis of the femur, the recorded values of the angle γ (part II, page 29) must be decreased by 8°.

The cervico-diaphyseal angle of each prosthesis is about 120° (part II, page 58), and if the angle α obtained from the prosthesis shall be compared with the opposite side or another hip-joint, the differences of the cervico-diaphyseal angles must be accounted for. Comparison of the two sides requires the angle α of the prosthetic side to be decreased 10° in case 1 and 20° in case 2.

The fit of the prosthesis in the acetabulum is necessary for good function. In case 1 the diameter of the prosthetic head was about 3 mm less than the diameter of the femoral head of the opposite side. The irregularities occurring during the recording of case 1 could be due to the dissimilarity of the diameters, or to minor movements of the prosthetic stem. In case 2 there was no measureable difference between the diameters of the prosthetic head and the femoral head replaced by the prosthesis.

In addition to the new conditions created by the prosthesis, damage secondary to the operation could have had an influence on normal function. A dorso-lateral approach was used in both cases. The gluteus maximus was divided and the small external rotators were cut. EMG-tests from the abductors, adductors and gluteus maximus and minimus showed no sign of neurogenic injury. Function of the short rotators was damaged to a certain extent. In case 2 there was a decreased ability to externally rotate.

Results show differences occurring between repeated tests and between the two patients. Some of the differences were due to the different ways of performing the tests. Furthermore the lever arms and the direction of muscle pull differs between the two patients.

The force acting on the prosthetic head in flexion, extension and abduction is large. In extension the difference between case 1 and case 2 is marked, but it is due to a different technique of extension. Case 2 rotated her pelvis when extending the hip. In flexion and extension the acting force exceeds the body-weight and in abduction it is about half the body-weight. Flexion, abduction and extension causes a force exceeding that of standing on two legs. While walking in a walking support the force is smaller. While sitting, no force exceeding 12 kp was recorded in either case. Of course while changing positions, for instance from a bed to a chair, the recorded forces were greater.

While flexing and abducting the opposite leg the forces acting on the prosthesis were not negligible (part IV, page 77, 80 and 81). Fig. 39 shows the force acting on the hip-joint when subjected to different loads due to shifting of the superimposed bodyweight. The test was performed with the patient standing with each leg on one of the walk-ways, shifting his weight from one leg to the other. How much of the body-weight was supported by respective legs was recorded from the walk-ways. The force acting on the hip when the walk-way recorded no force under the

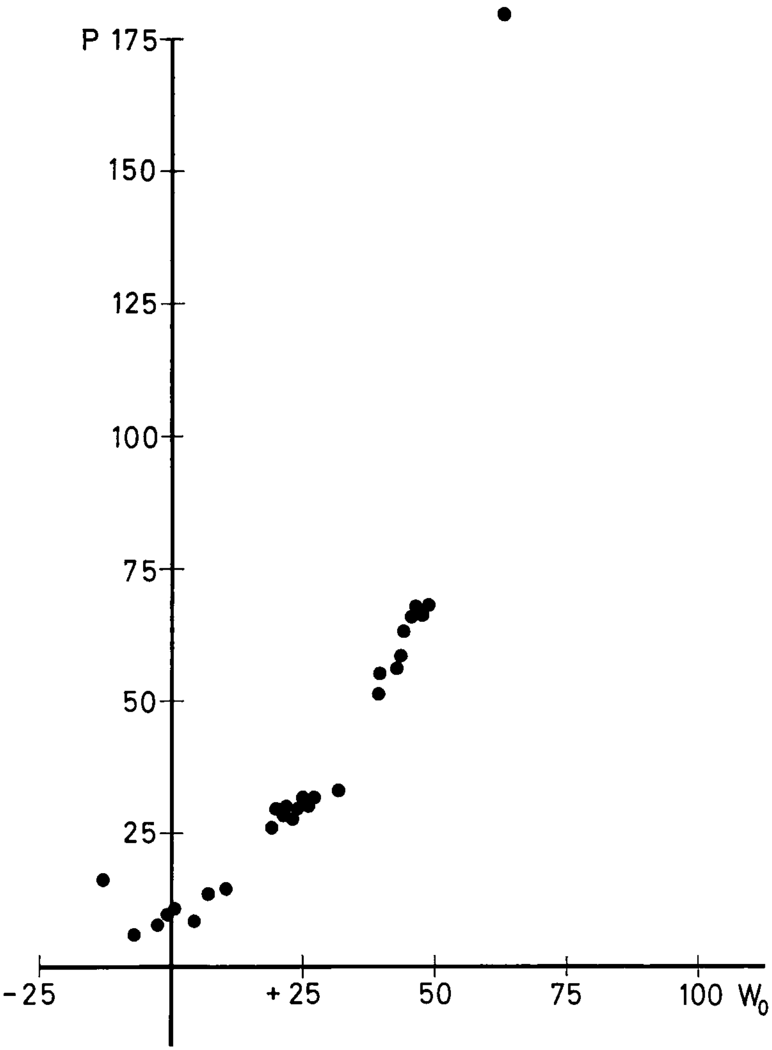

Fig. 39. *The force acting on the prosthetic head when subjected to different loads due to shifting of the superimposed body weight. The weight of each leg is approximated to* $^1/_6$ *of the body weight. The force P is indicated on the y-axis and the superimposed body weight* W_0 *on the x-axis. The maximum value is for one leg support on the prosthetic leg and the negative x values are for one leg support of the opposite side.*

prosthetic leg was 0.3 times the body-weight. The discrepancy between this value and that of one-leg support of the opposite leg (part IV, page 86) is due to flexion of the knee in the latter test.
The force acting in two-leg support was close to 0.5 times the superimposed body-weight or about 1/3 of the body-weight. One-leg support gives rise to a high load on the hip. One-leg support of the opposite side especially in case 2, gives a surprisingly high load on the hip. In case 2 this developed with about 90° of flexion in the hip. One-leg support results in a purely vertical load, parallel to the longitudinal axis of the femoral shaft.
While walking, the hip must resist great forces in both the stance- and the swing-phase. In the stance-phase two maximum points with a low point between occur similarly to the vertical force curve describing the action between the foot and the ground. At the transition between stance- and swing-phase the force in the hip is not zero as the vertical force F is. In the swing-phase two maximum points with a low point between occurred for case 1. The first maximum point appears immediately after toe-off, and the second immediately before heel-strike. The force at the low point was about zero.
In case 2 the transition between the stance- and the swing-phase did not cause any changes in the curves, except an occasional decrease in the slope of the curves. In the middle of the swing-phase a low point, Mn, occurred and the forces at this point were about zero. The maximum point Mx_2, which occurs in the swing-phase and the low point *a*, which occurs at the transition between the swing- and stance-phase in case 1 were represented by a plateau in the curves. Occasionally, however, a maximum point was followed by a low point similar to case 1.
While walking down-stairs (part IV, page 100) the shape of the curves changed in the swing-phase. There were two low points recorded and between them a high point. Forces present during the swing-phase were smaller than those in the stance-phase but sometimes reached values approximating body-weight.
In case 1 walking tests were performed at two walking-speeds. The forces present during faster walking were greater, as was expected. With a walking-speed of 0.9 m/sec. the maximum force was about 1.59 times body-weight in the stance-phase and 0.59 times body-weight in the swing-phase. At the speed of 1.3 m/sec. these forces were 1.80 and 0.85 times body-weight respectively.
In case 2 the walking tests were performed similarly to those for case 1. With a walking speed of 1.1 m/sec, the maximum force was about 3.0 times body-weight in the stance-phase and 1.2 times body weight in the

swing-phase. At the speed of 1.4 m/sec., these forces were 3.3 and 1.2 times body-weight respectively.

In case 2, the tests also included running, which created forces of 4.3 times body-weight in the stance-phase and 2.5 times body-weight in the swing-phase. Running-speed varied slightly and each speed gave about the same results. The maximal recorded value was slightly more than 4 times body-weight. Very slow walking caused the forces in the stance-phase to slightly exceed body-weight. Those forces of the swing-phase were negligible. With use of walking-supports, the force could be reduced to about 0.3 times the body-weight.

A study of cane-walking showed that forces in the hip were reduced most when using the cane in the opposite hand. If the cane was used in the

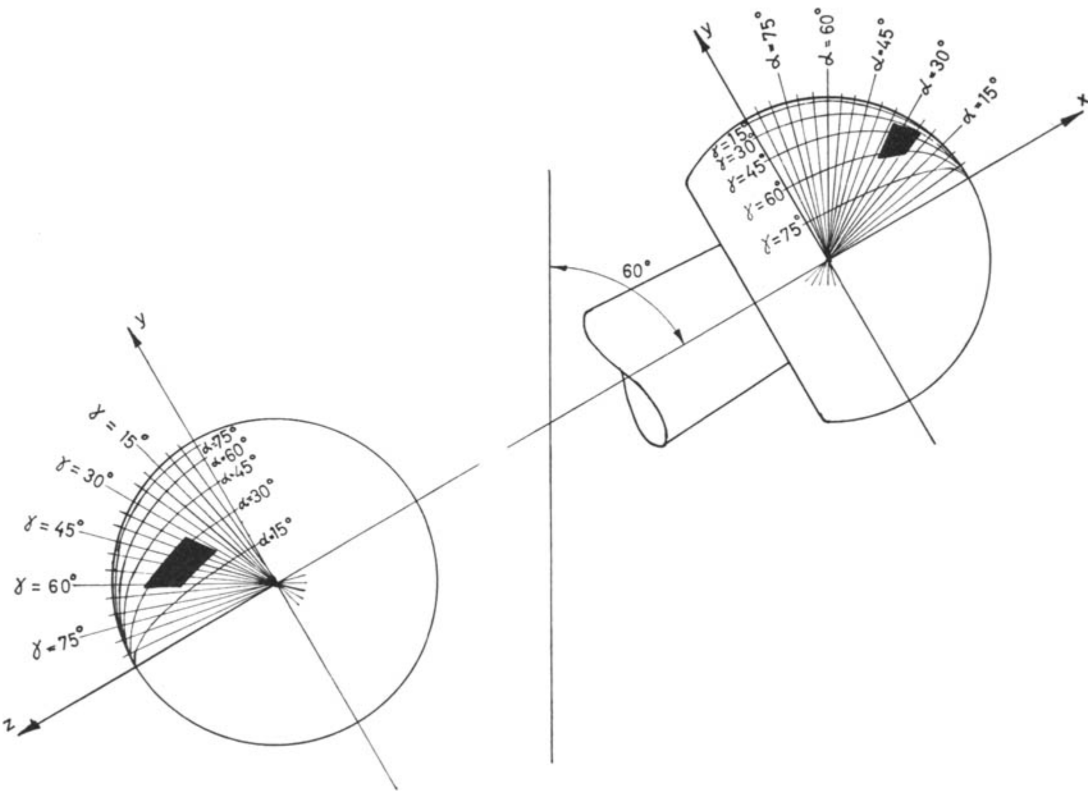

Fig. 40. *Case 1. Flexion, extension and abduction. Two views of the prosthetic head, each of which is graduated. The right view shows the ventral surface of the head as seen from an anterio-posterior direction and to the left the surface of the head as seen from acetabulum. The area of application of the resultant force, as derived from the meanvalue and standard deviation of* $\alpha = 28^{\circ} \pm 5^{\circ}$ *and* $\gamma = 47^{\circ} \pm 15^{\circ}$*, is indicated.* $n=21$

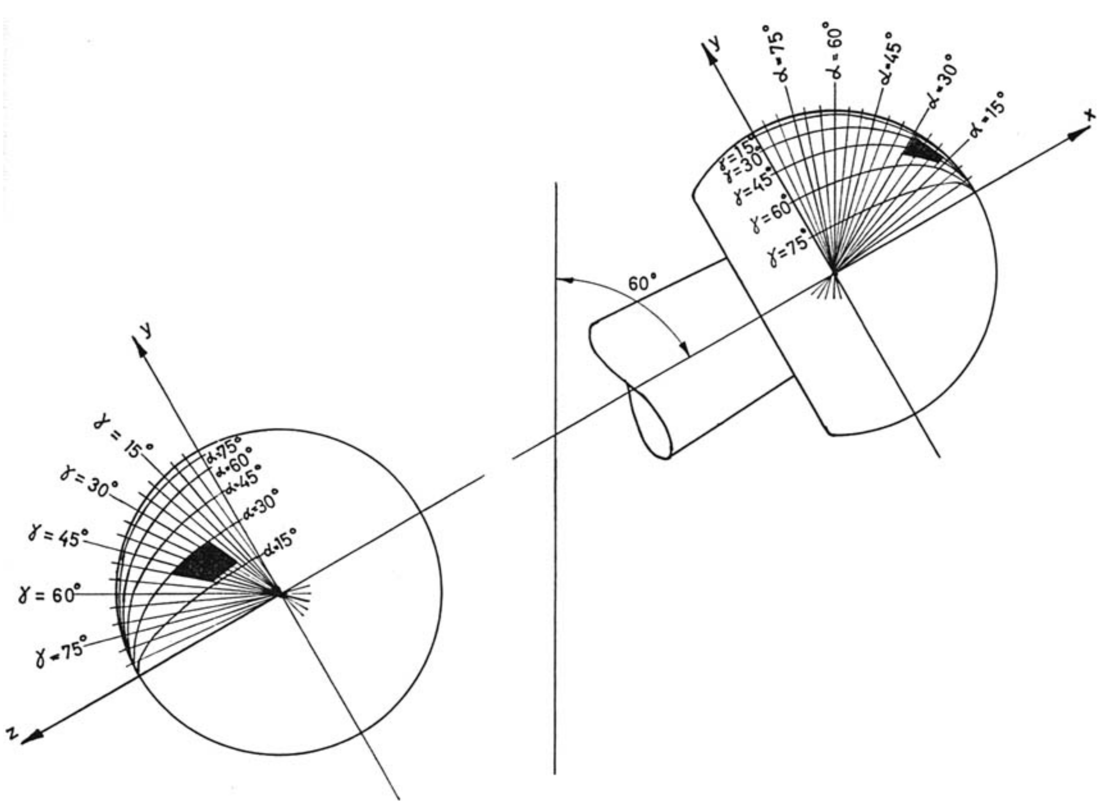

Fig. 41. *Case 2. Flexion, extension and abduction. Two views of the prosthetic head, each of which is graduated. The right view shows the ventral surface of the head as seen from an anterio-posterior direction and to the left the surface of the head as seen from acetabulum. The area of application of the resultant force, as derived from the mean value and standard deviation of $\alpha = 23^\circ \pm 6^\circ$ and $\gamma = 37^\circ \pm 12^\circ$, is indicated. $n=31$*

prosthetic hand, the vertical force between the foot and the ground was reduced on the unoperated side.

The coefficient of friction between the steel surface of the prosthetic head and the acetabular cartilage was found to be low, about 0.02—0.04. This is more than double that between cartilage and cartilage but less than that occurring between two lubricated steel surfaces. In case 1 the coefficient of friction increased somewhat after walking about 25 m. At that point the patient complained of fatigue in the hip. This vanished after a few minutes' rest and the coefficient of friction returned to its original lower value. These findings could not be verified in case 2.

In case 1 the coefficient of friction was also determined under static conditions. It was found to be about 0.10 which is markedly more than that for dynamic conditions.

Traction up to 24 kp for 3 hours applied to the prosthetic leg caused no forces acting on the prosthetic head.

Tests performed on case 1 and 2 on regaining contcionsness after anaesthesia revealed no force to act in a relaxed supine position.

An interesting observation has been made regarding the direction of the acting forces. The area of application of the resultant force is very limited regardless of the head's position in the joint. This means, that the direction of the force is constant in relation to the prosthetic head. The area of application is not the same as the contact or compression area, but since it is applied between two smooth surfaces it must be the midpoint of the contact area. If the surfaces are not congruent, this becomes an area of

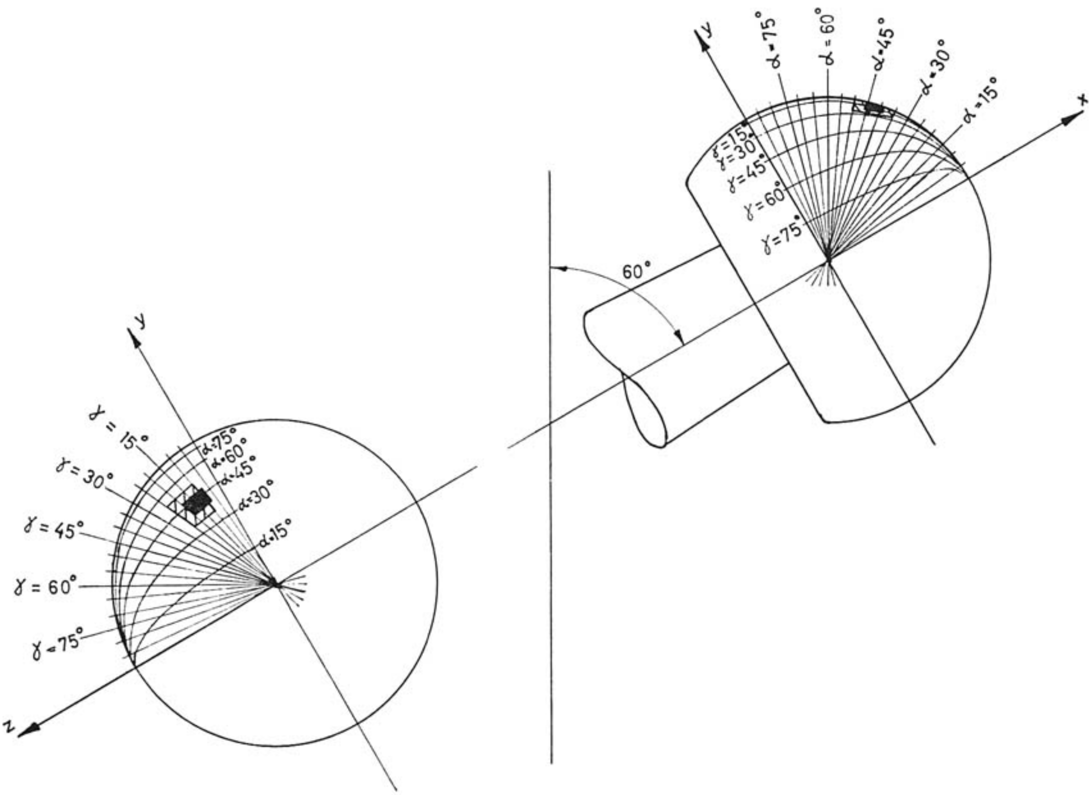

Fig. 42. *Case 1. Level walking. Two views of the prosthetic head each of which is graduated. The right view shows the ventral surface of the head as seen from an anterio-posterior direction and to the left the surface of the head as seen from acetabulum. The area of application of the resultant force, as derived from the mean value and standard deviation of* α *and* γ *is indicated. The solid black area represents the values* $\alpha = 43^\circ \pm 3^\circ$ *and* $\gamma = 15^\circ \pm 5^\circ$ *in the stance-phase and the striped area the values* $\alpha = 43^\circ \pm 8^\circ$ *and* $\gamma = 19^\circ \pm 6^\circ$ *in the swing-phase. n=99 for the stance — and n=66 for the swing-phase.*

algebraic summation of points resulting from two or more contact areas. It is hard to believe that there can be more than one contact area at the same time between the prosthetic head and the acetabulum, and this study indicates that the area of application is the centre of the contact area. In fig. 40 and 41 the area of application for all tests in flexion, extension and abduction are illustrated. This area is derived from a mean value and its standard deviation.

In fig. 42 and 43 the area of application is derived the same way and illustrates data from the tests of level walking in case 1 and level walking and running in case 2. Values have been taken from all maximum points in the stance- and swing-phases and the point *c*; in running also for the

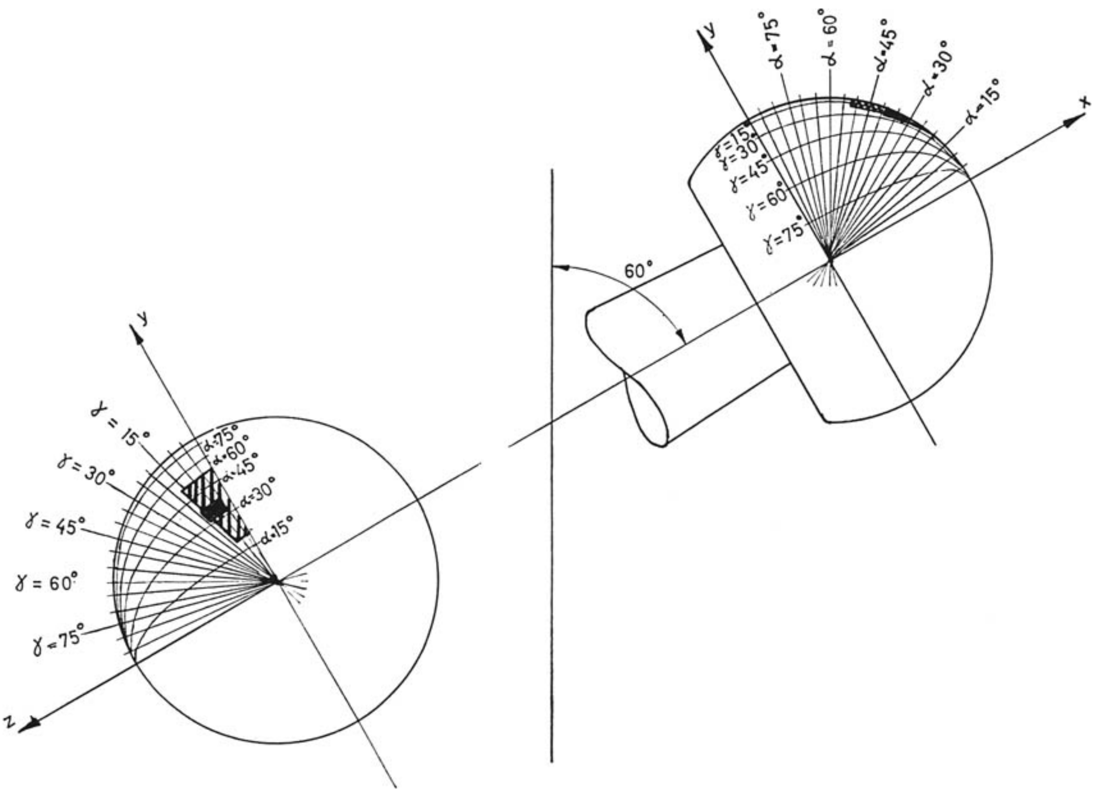

Fig. 43. *Case 2. Level walking. Two views of the prosthetic head each of which is graduated. The right view shows the ventral surface of the head as seen from an anterio-posterior direction and to the left the surface of the head as seen from acetabulum. The area of application of the resultant force, as derived from the mean value and standard deviation of* α *and* γ *is indicated. The solid black area represents the values* $\alpha = 3>^\circ \pm 3^\circ$ *and* $\gamma = 11^\circ \pm 7^\circ$ *in the stance-phase and the striped area the values* $\alpha = 36^\circ \pm 17^\circ$ *and* $\gamma = 9^\circ \pm 8^\circ$ *in the swing-phase.* $n=120$ *for the stance — and* $n=40$ *for the swing-phase.*

points *a* and *e*. The area of application is small and differs little from that of flexion, extension and abduction.

In fig. 44 and 45 the area of application of the force present in stair-walking is shown. This area is somewhat larger but involves the usual part of the head.

In addition to the force resultant being applied to a limited area on the prosthetic head, it was observed that the horizontal component was always applied to the head on its ventral side. With exception of the low points which occurred in the swing-phase, the force was never applied to the head on its dorsal aspect when the patient actively moved the hip. To check these results, rotation tests were made in case 2. Active internal

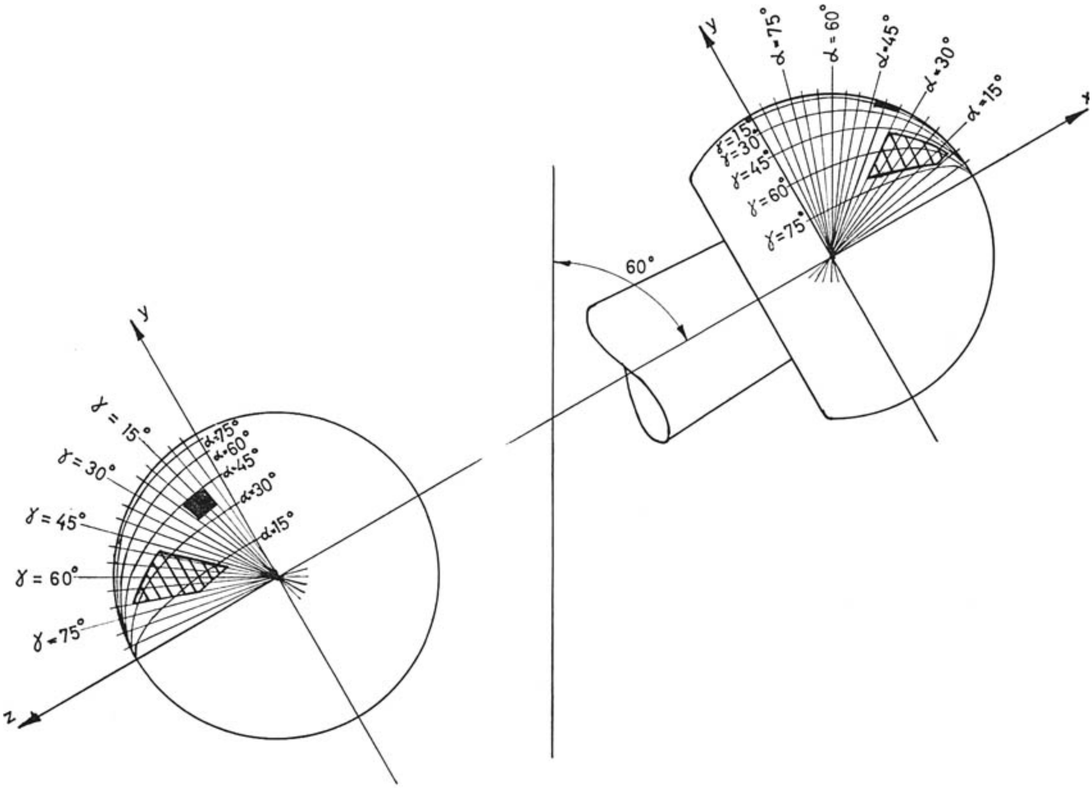

Fig. 44. *Case 1. Stairwalking. Two views of the prosthetic head each of which is graduated. The right view shows the ventral surface of the head as seen from an anterio-posterior direction and to the left the surface of the head as seen from acetabulum. The area of application of the resultant force, as derived from the mean value and standard deviation of* α *and* γ *is indicated. The solid black area represents the values* $\alpha = 39^\circ \pm 4^\circ$ *and* $\gamma = 18^\circ \pm 5^\circ$ *in the stance-phase and the striped area the values* $\alpha = 23^\circ \pm 11^\circ$ *and* $\gamma = 59^\circ \pm 11^\circ$ *in the swing-phase. n=51.*

rotation should cause the head to be pressed against the dorsal part of the acetabulum. However, when the patient actively rotated the thigh inwards the force was always applied to the head on its ventral side. This occurred whether lying in bed with the hip and knee in the neutral position or while sitting in a chair with hip and knee flexed. If the patient was told to relax, passive internal rotation created a force which now was applied to the head on its dorsal side. From this it is concluded that moving the hip actively causes the force always to come from the ventral side. With the hip flexed while supine the force had the same direction as in extension while prone.

The limited direction of the force in relation to the prosthetic head surface

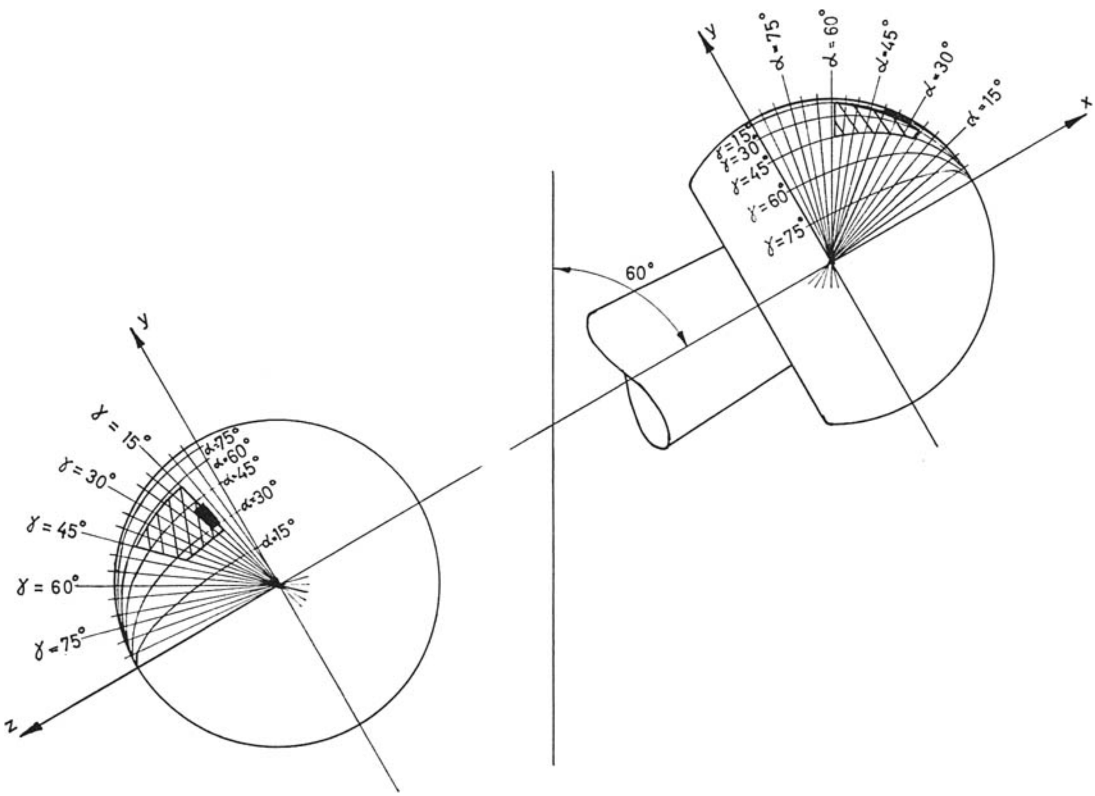

Fig. 45. *Case 2. Stairwalking. Two views of the prosthetic head each of which is graduated. The right view shows the ventral surface of the head as seen from an anterio-posterior direction and to the left the surface of the head as seen from acetabulum. The area of application of the resultant force, as derived from the mean value and standard deviation of* α *and* γ *is indicated. The solid black area represents the values* $\alpha = 36^\circ \pm 5^\circ$ *and* $\gamma = 15^\circ \pm 2^\circ$ *in the stance-phase and the striped area the values* $\alpha = 42^\circ \pm 15^\circ$ *and* $\gamma = 30^\circ \pm 15^\circ$ *in the swing-phase. n=50.*

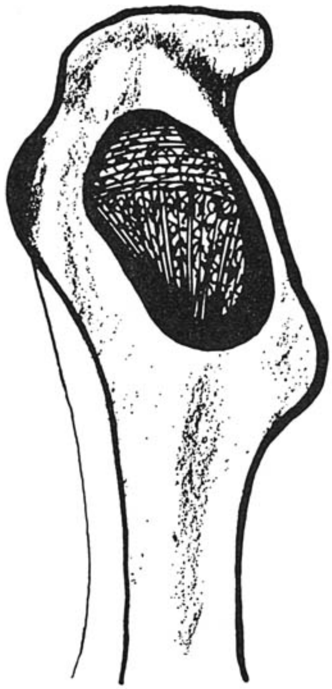

Fig. 46. *In a cross section through the junction of the femoral neck and shaft, the lamellae assume the shape of a T-beam.*

is in accordance with some details of the anatomy of the upper femur and also with findings in some pathological conditions associated with decreased resistance against mechanical factors. The femoral neck is not a true cylinder but reveals an elliptical cross-section. This elliptical shape is less pronounced near the head, but increases in distal-lateral direction. The major axis of the elliptical cross-section forms an angle opening dorsally relative to the shaft. This gives maximum resistance to a force parallel with the major axis. Such a force will be applied to the femoral head on its ventral aspects. Furthermore the structural resistance increases in the distal-lateral direction to counterbalance the stresses present in this type of cantilever.

It is interesting to note that a cross-section taken through the neck close to the shaft reveals the lamellae of the spongiosa to run in such a way that its design is in the form of a T (fig. 46). This could mean that the spongiosa at this level acts as a T-beam to give high resistance to vertical loading. Since the base of the T comes from the lateral trabecular portion and the upper part of the T from the arcuate portion, this could be one explanation of the function of the two systems. The upper part of the T will resist tensile stresses and the lower part compressive stresses. The

direction of each part of the T coincides approximately with the direction of one of the two axes of the cross-sectional ellipse.
The antetorsion usually seen in the upper femur presents a high resistance to any horizontal force applied to the head on its ventral aspects.
Clinically, the presence of a ventrally acting force may help to explain why femoral head displacement is always dorsal in femoral neck fractures and in slipped capital femoral epiphyses.
In some pathological conditions where tissue alteration occurs secondary to mechanical factors, X-ray changes are often seen in an area corresponding to the force area of application.
In coxa plana indentation and fragmentation are usually found in this area. (Edgren 1965).
Trueta et al (1953, 1954) described an area in which they found pathological changes in osteoarthritis of the hip. This area, which they called the compression area, is described by a U-shape and covers the upper ventral and dorsal aspects of the femoral head.
A review of X-rays of osteoarthritis in the hip-joint shows the first sign of indentation occurring on the ventral side of the longitudinal axis of the femoral neck in the lateral views.
One of the first signs in X-rays where necrosis of the femoral head is present usually is a compression in the circular outline of the head in the same area as the area of application of the force. Examination of excised necrotic femoral heads usually shows a compressed area most pronounced in the predicted area.

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