The morphology and functions of the muscles around the hip joint after a unilateral transfemoral amputation

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SUMMARY

THE MORPHOLOGY AND FUNCTIONS OF THE MUSCLES AROUND THE HIP JOINT AFTER A UNILATERAL TRANSFEMORAL AMPUTATION

This dissertation is concerned with the consequences of a transfemoral amputation for the morphology and functions of the muscles around the hip joint. Knowledge about and insight into the changes appearing in the morphology and functions of the hip muscles of transfemoral amputees are important to amputation surgery as well as to prosthetics. Amputation surgery and prosthetics are interrelated and cannot be separated one from the other. The amputation should be performed in such a manner that a good functional stump will be achieved. The quality of the stump is determined by the level of amputation, the amputation method, the method(s) of muscle reattachment and the final shape of the stump. The prosthesis should be designed in such a way that the stump can act functionally in transferring forces to and from the prosthesis and the ground. The quality of the prosthesis is determined by the design and fit of the socket and the components and alignment of the prosthesis. In order to improve and optimize amputation surgery and prosthetics and the collaboration between both disciplines more knowledge about and insight into the morphology and functions of the hip muscles of transfemoral amputees is necessary. The aims and theoretical principles of this study are discussed in chapter 1.

Chapter 2 describes the method designed to determine the geometry and volumes of the bones and muscles in the hip and thigh region. The method is based on three-dimensional (3D) reconstruction of contour-outlined two-dimensional (2D) Magnetic Resonance Images (MRI). The contours of the bones and muscles are traced manually. First the cross-sectional areas of the traced contours of a structure are calculated, after which the volume of the structure is determined as the sum of the products of the cross-sectional areas times the slice thickness (5.5 mm).

In chapter 2 also the morphology of the hip muscles of three normal subjects has been studied, for one thing to determine the reliability

and reproducibility of the method and for another as reference material for the study of the changes appearing in the morphology of the hip muscles after a transfemoral amputation.

The reliability and reproducibility depend on the accuracy with which the contours of the structures have been traced. Inaccuracy in tracing and thus in the volume calculations can be due to technical factors, such as the gray level variation and frequency artifacts in the MR images (especially between muscular and adipose tissue) and subjective factors, such as the subjective interpretation of the contours of the structures in the MR images (interpretational error). The relative error in the volume calculations (%) due to technical factors depends on the cross-sectional area of the contours and decreases with increasing cross-sectional area; 10% for an area of 125 mm^2 up to 0.1% for areas larger than 500 mm^2. The interpretational error depends, apart from the size of the cross-sectional area, on the visibility of the contours. Bones are clearly visible, just like muscles embedded in fat or surrounded by a thick fascia. The interpretational error of these structures is low (0-3%). Muscles which are closely apposed to each other without a thick intervening fascia (the three vasti separately, adductor minimus and brevis) are difficult to separate in the images, resulting in a high interpretational error (10%).

The cross-sectional areas of all muscles in the hip and thigh region are much larger than 500 mm^2, as a result of which the errors of tracing and of the volume calculations are relatively low (0-3%). This does not apply to the tendons, which have small cross-sectional areas. MRI in combination with 3D-reconstruction accordingly provides an accurate and detailed anatomical description of the geometry and volumes of the bones and muscles in the hip and thigh region. MRI gives the possibility of studying superficial as well as deep muscles under tension in the living state.

On the basis of the designed method the morphology of the bones and muscles in the hip and thigh region has been studied in twelve male unilateral transfemoral amputees. The bones and muscles at the amputated side change considerably after a transfemoral amputation. In ten amputees atrophy of the substantia compacta of the femur has occurred (10-40%); the marrow cavity has increased in volume and the thickness of the compact bone of the femur has decreased.

The partly amputated muscles (40-60%) as well as the intact muscles at the amputated side (0-30%) are atrophied. Cleaved muscles, which are not reanchored, are retracted and show fatty degeneration. However, if the partly amputated muscles have been reanchored correctly, these muscles are atrophied but not degenerated into fat. The amount of atrophy of the intact muscles (iliopsoas, gluteus medius and minimus) is statistically significant dependent on the stump length. The crucial decrease in muscle volume appears in amputations in the proximal half of the femur.

Also, the geometry of the cleaved and reanchored muscles changes (sartorius, rectus femoris, tensor fasciae latae, hamstring muscles and gracilis), which is due to a changed distal fixation and which has consequences for the function(s) of the muscle. Especially in amputations in the proximal half of the femur a changed geometry of the once biarticular muscles is visible.

Muscles reanchored by a myoplasty (myoplasty is the end-to-end suture of antagonistic muscles or muscle groups across the cut end of the femur) show a greater volume than muscles reanchored by a myodesis (myodesis is the attachment of the muscles to the femoral stump). The differences in muscle volume are due to the fact that muscles reanchored by a myodesis only have a smaller cross-section at the level of fixation to the femur in contrast to muscles reanchored by a myoplasty.

In seven transfemoral amputees the iliotibial tract has not been reanchored, probably to avoid an abduction contracture. Nevertheless in four of these amputees, all with a short or medium stump length, an abduction contracture has appeared. In two of these amputees the cleaved adductor longus and magnus have also not been reanchored. These muscles are strong hip adductors because of their distal insertion at the femur. The hip abductors remain intact after an amputation so that there is no balance in muscle force between the abductors and adductors, resulting in an abduction contracture of the femur.

The case for the hip flexors and extensors is similar. If the iliotibial tract is not reanchored the tensor fasciae latae and gluteus maximus are retracted, because both muscles insert into the iliotibial tract. The gluteus maximus is the most important hip extensor in transfemoral amputees, because the hamstring muscles are always affected by the amputation. If the gluteus maximus is retracted and has no insertion, the muscle is less effective. The strongest hip flexor, the iliopsoas, remains intact in all amputations. This results in an imbalance of muscle force between the hip flexors and extensors and a flexion contracture of the femur appears. Four amputees with a short or medium stump length show a flexion contracture.

The conclusion of the study described in chapter 3 is that a balance of muscle force between the hip adductors and abductors and between the hip flexors and extensors is important to avoid a contracture and can only be obtained if all the cleaved muscles and the iliotibial tract are reanchored correctly, with the right muscle tension and with the femur in a neutral position.

In chapter 4 the maximal isometric torque (MIT) of the hip muscles in four different directions (flexion, extension, adduction and abduction) has been determined. The results of this study are compared with the morphological results of chapter 3. Twelve males without an amputation have been studied as a reference group.

The MIT at the intact side of the amputees is equal to the MIT of the subjects without an amputation. The MIT at the amputated side is on average lower than the MIT at the intact side of the amputees; flexion is 32%, extension 35%, adduction 50% and abduction 46% lower. The differences are statistically significant, except for the hip extension. High correlations (0.76-0.92) are found between the reduction in MIT (%) and the amount of atrophy (%) of the hip flexors, extensors and abductors but not for the hip adductors (0.64). Also a correlation is found between the reduction in muscle torque and the stump length (0.68-0.87).

In amputations in the distal third of the femur the maximal flexion, extension and abduction torques decrease 10-20% and the maximal adduction torque 30-40%. In amputation around the middle of the femur the maximal flexion and extension torques decrease 30-40% and the maximal abduction and adduction torques 40-50%. In amputations in the proximal third of the femur the maximal isometric torques decrease about 70-80%. The crucial decrease in muscles torque, in agreement with the crucial decrease in muscle volume, appears in amputations in the proximal half of the femur. Especially in amputations in the proximal half of the femur it is important to keep the stump as long as possible to retain maximum force of the hip muscles.

Amputees with an abduction contracture show a substantially lower maximal abduction torque, which indicates that the abductors are in a permanently shortened state. From the literature it is known that the maximal isometric abduction torque decreases with progressive hip abduction.

In chapter 5 the kinematic parameters of the prosthetic gait and the activity of the superficial hip muscles at the intact and amputated side have been described in the same transfemoral amputees. The gait of the transfemoral amputees with a prosthesis has changed. The amputees find problems in walking with a prosthesis, which are due to the passive element of the prosthetic leg. The hip muscles are not directly connected with the prosthesis, as a result of which forces have to be transferred to and from the prosthesis to move the prosthetic leg. Also, rotations in the prosthetic knee have to be initiated by the muscles of the hip and thigh.

The prosthetic gait shows some remarkable characteristics.

The amputees walk with a 35% lower comfortable walking speed than normal subjects. However, there was a large variation in comfortable walking speed (0.71-1.30 m/s) between the amputees, but this speed is not correlated with the age or stump length of the amputees. The stride length also shows a large variation, but the step rate a small variation. It seems that the amputees compensate for the lower or higher walking speed with their stride length rather than with their step rate.

The amputees walk with a larger stride width (18-30 cm) than normal subjects (16 cm). The amputees with a high comfortable walking speed walk with a smaller stride width than amputees with a low comfortable walking speed.

The amputees show an asymmetrical walking pattern. In normal subjects the stance phase takes approximately 58% of the stride and the swing phase approximately 42% and these values are equal for both legs. However, in the amputees the durations of the stance and swing phases are not equal for both legs. The amputees stand a little longer on their intact leg (63%) than on their prosthetic leg (58%).

In normal subjects there is only little lateral bending of the trunk and this is equal for both sides (3-5 degrees). Eight out of eleven amputees show a stronger lateral bending of the trunk toward the prosthetic side during the stance phase of the prosthetic leg (7-25 degrees). From the literature it is well known that there is a relationship between lateral bending of the trunk and insufficient function

of the hip abductors. In four amputees with extreme lateral bending of the trunk the hip abductors (gluteus maximus and medius) are indeed strongly atrophied (chapter 3) and these amputees also show a lower maximal isometric abduction torque (chapter 4). These amputees all have an abduction contracture and their iliotibial tract is not fixed, resulting in a retracted gluteus maximus.
These findings confirm the importance of a good fixation of the adductor longus and magnus and the iliotibial tract to avoid an abduction contracture and to obtain a well functioning gluteus maximus. In this manner a good function of the hip abductors will be obtained, resulting in a sufficient stabilization of the pelvis which is an important condition for a good walking pattern.
The patterns of motion of the hip and knee joints also show some remarkable characteristics. In contrast to normal subjects, in the amputees rebound of the hip at the amputated side at heel strike of the prosthetic foot is very small or absent. The absence of the rebound of the hip is connected to the absence of a rebound of the prosthetic knee at heel strike as in normal subjects. The intact knee of the amputees flexes a few degrees at heel strike, but remains in a flexed position during the entire stance phase in contrast to normal subjects.
Some amputees show a larger hip flexion at the end of the swing phase than normal subjects, probably to increase the swing velocity of the knee to ensure a stable extended knee.
The amputees also show a larger hip extension at the end of the stance phase, except amputees with a very short stump. Normal subjects flex their knee gradually during hip extension (knee flexors and extensors) and unroll their foot at the same time (plantar flexors). However, in the prosthetic leg these active mechanisms are absent and these movements have to be initiated by the pelvis and thigh. The prosthetic knee is only bent shortly before toe off, which results in an increment of the hip extension.
In amputees with a short and medium stump length the angular velocity at the end of the hip extension increases, immediately followed by a fast hip flexion, as a result of which the hip flexion-extension curve shows a sharp transition from hip extension to flexion. Amputees with a short or medium stump length only have a small contact area between the thigh and socket to transfer forces to and from the prosthesis.

Consequently the hip flexors are hardly able to swing the prosthetic leg forwards and the hip flexion has to be initiated by the muscles of the trunk and pelvis. Shortly before the swing phase the amputees tilt their pelvis and trunk forwards, followed by a fast backward tilting of the pelvis, resulting in a fast forward swing of the thigh (hip flexion).
The activity of the superficial hip muscles during walking has also been studied in this chapter. The intact hip muscles of the amputees show the same phasic activity during a stride as those of normal subjects, but during a longer period.
The activity of the gluteus maximus and medius at the amputated side during the stance phase increases with decreasing stump length. These muscles are important hip stabilizers. The shorter the stump length the more problems the amputee finds in stabilizing the pelvis and the prosthetic leg during the stance phase. This is due to the fact that the femur is not directly in contact with the floor. In normal subjects the femur has a stable support on the floor during the stance phase, as a result of which the hip stabilizers have adequate leverage. In the amputees the femur is incorporated into the socket and has no stable position. When the gluteus maximus and medius are active to stabilize the pelvis the femur is not fixed immediately, but must first compress soft tissues of the lateral thigh, before the femur is stabilized against the wall of the socket.
The longer period of activity during the stance phase is evident in all hip muscles at the amputated side.
The phasic action of the once biarticular muscles during a stride has changed the most, which is due to the absence of the knee joint. In amputees with a long stump the once biarticular muscles are still active during the swing phase (flexors) or at the beginning of the stance phase (extensors). In amputations in the proximal half of the femur these muscles are continuously active. Probably these muscles play a role in fixing the socket.
The extra movements and muscle activity of the amputees performed in walking with a prosthesis will probably cause a higher energy expenditure. In chapter 6 the energy expenditure of eleven male transfemoral amputees and six normal males has been measured at different walking speeds.
First a test protocol was designed to determine the energy expenditure

at different walking speeds, because available tests were not suitable for transfemoral amputees with a prosthesis. From the literature it is known that in normal subjects the walking speed experienced as most comfortable is very close to the most metabolically efficient walking speed. It is not known if this applies equally to transfemoral amputees. In this chapter the comfortable and most metabolically efficient walking speeds of normal subjects and transfemoral amputees have been determined. The most metabolically efficient walking speed is the speed with the least energy expenditure per distance covered.

To determine the most metabolically efficient walking speed the energy expenditure was determined at six incremental walking speeds with intervals of 0.14 m/s (0.5 km/h). Between successive walking speeds the subject rested during three minutes. During walking also the heart rate and step rate were determined; during rest only the heart rate. The measurements were carried out on a treadmill. Not all the amputees in this study were able to walk without the help of a walking stick or the bars of the treadmill, especially at the higher walking speeds. Using a walking stick or the bars probably alters the energy expenditure. Therefore these amputees were combined in a separate group (group 2).

The mean comfortable walking speed of the normal subjects is 1.36 m/s, of group 1 of the amputees 0.83 m/s and of group 2 of the amputees 0.74 m/s. If the amputees walk with their own comfortable walking speed a 25-35% higher energy expenditure is found in comparison with normal subjects at the same speed. The difference in energy expenditure increases with the walking speed.

Also the heart rate of both groups of amputees is higher than that of normal subjects, but this difference is not statistically significant. If the normal subjects as well as the amputees walk at their most metabolically efficient walking speed their energy expenditure per second is equal. However, the amputees walk with a lower most metabolically efficient walking speed than normal subjects. The mean most metabolically efficient walking speed of the normal subjects is 1.30 m/s and of group 1 and 2 of the amputees 0.94 m/s. This implies that the comfortable walking speed of the amputees does not correspond with the most metabolically efficient walking speed.

At the comfortable walking speed the amputees have the same step rate as the normal subjects. In the amputees the step rate increases less

with the walking speed than in normal subjects. When the amputees walk with a higher or lower walking speed they compensate for the more or less constant step rate with a different step length. This is found also in chapter 5. The more or less constant step rate may be connected with the fact that the amputees have difficulty controlling the swing speed of the knee mechanism. The function of the prosthesis as an externally driven pendulum influences the spontaneous choice of step rate.
From the literature it is known that each walking speed has a most metabolically efficient step rate and that normal subjects spontaneously choose the efficient step rate. If the normal subjects walk at a deviating step rate their energy expenditure increases. If the amputees walk at their comfortable walking speed the step rate is in agreement with the walking speed. However, at a higher or lower walking speed the step rate differs from that of normal subjects. Probably a part of the higher energy expenditure of the amputees is due to their deviating step rate.
The findings of this study show the importance of a proper surgical amputation. Accurate fixation of all the cleaved muscles, but especially of the adductor longus and magnus and the iliotibial tract, is important to avoid contractures, as a result of which a good functioning stump will be obtained. The prosthetist will encounter fewer problems in designing and creating a good functioning prosthesis if the amputee has a strong stump without contractures. The more deviations the stump has the higher the demands put on the prosthetist and the prosthesis to obtain a good walking pattern. A good functioning stump and prosthesis are important conditions to achieve a good walking pattern with energy expenditure as small as possible.
This study gives more insight into the morphology and functions of the intact and cleaved hip muscles of transfemoral amputees. The new viewpoints will hopefully lead to new principles to improve and optimize amputation surgery as well as the supply of prostheses to patients.