Transfemoral Amputees Walking on a Rotary Hydraulic Prosthetic Knee Mechanism: A Preliminary Report

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ABSTRACT

Results from multiple instrumented gait analysis trials of seven traumatic transfemoral amputees capable of community ambulation are reviewed. All subjects used a novel rotary hydraulic prosthetic knee offering both stance and swing phase control, in combination with various contemporary prosthetic ankle-foot mechanisms. Unlike previous hydraulic stance and swing units, the Otto Bock 3R80 knee provides stance stability whenever the desired weightbearing load is applied to the prosthesis.

The mechanical principles and basic functions of the knee joint are described. Results from three phases of gait studies are presented to identify inter-individual variability and the effect of varying sagittal plane knee alignment and the ankle-foot mechanism selected. Subjective amputee preferences for specific foot devices are presented.

The pattern of hip moments on the amputated side during stance phase are characteristic of the individual amputee. However, the magnitude and timing of the hip moments vary considerably between individual amputees. So long as the prosthetic knee remains biomechanically stable, it appears that alignment alterations or the use of varying foot devices have little impact on the hip extension moment. On the other hand, the hip flexion effort required for the amputee to initiate knee flexion in late stance phase is directly related to the linear position of the prosthetic knee in the sagittal plane: the more posterior the knee, the greater the effort needed to begin swing phase.

Clinical function of this hydraulic prosthetic knee was considered satisfactory with all tested ankle-foot mechanisms. Transfemoral amputee preference for specific prosthetic feet in this study seemed to be determined by two biomechanical factors: (1) their ability to benefit from the use of a dynamic response foot without compensating with the non-amputated knee, and (2) their ability to generate a hip extension moment on the prosthetic side during the weight acceptance phase of gait.

Key Words: Biomechanics, Gait Analysis, Prosthetic Knee Joint, Prosthetic Fitting, Leg Prosthesis.

Introduction

For many decades, the Mauch S-N-S Hydraulic cylinder was the only compact fluid control system for prosthetic knees that offered integrated hydraulic stance and swing phase control in one unit.[1] The more recent CaTech cylinders function in a similar fashion. Elimination of the stance phase flexion resistance of these devices, which is much higher than is desirable for swing phase function, occurs whenever two conditions are met simultaneously: (1) the knee is in full extension, and the piston is therefore at the limit of its travel; and (2) a knee hyperextension moment (or tensile force on the piston) is applied for at least one tenth of one second.

During the gait cycle, this pattern typically occurs during terminal stance, but it may also sometimes happen at undesirable times, such as during the period of loading response.

The Otto Bock 3R80 knee joint is an innovative rotary design offering load dependent, or weightbearing-activated, hydraulic stance stability in addition to hydraulic swing phase control.[2,3] In contrast to previous piston-based stance-and-swing control units, when using the 3R80 the transition from stance to swing phase function (and vice versa) is not dependent on extension of the knee. The 3R80 also offers unique mechanical and hydraulically damped stance flexion capabilities.

The primary purpose of this article is to present preliminary gait data from amputee trials with this knee mechanism in combination with a variety of prosthetic feet. The effect of alignment changes on objective gait measurements will also be discussed. Familiarity with the mechanical capabilities of the 3R80 prosthetic knee is prerequisite for evaluating and understanding the gait analysis results presented later in this report.

Mechanical Principles of the 3R80 Prosthetic Knee

Basic Components:

A schematic overview of the 3R80 is presented in Figure 1 . The rotary axis of this single axis knee joint is also the rotary axis of the piston vane dividing the hydraulic chamber into two portions, which control extension and flexion movements, respectively. The chambers are connected by two channels containing one-way flow valves and individually adjustable resistance valves.

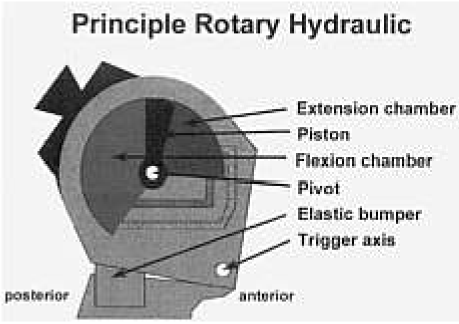


Figure 1. Schematic of rotary hydraulic knee.

The secondary trigger axis allows the joint mechanism to rotate into approximately four degrees of stance flexion. In addition to a fixed extension stop, there is a compressible elastic bumper resisting this mechanical stance flexion movement.

Functional Elements

Figure 2 depicts the functional elements influencing the amount of resistance to movement of the hydraulic oil, which fills the chamber and channels. During knee flexion the oil flows in a counterclockwise direction; the flow is reversed during extension movements. For the knee to rotate about the trigger axis and close the stance flexion resistance valve, the weightbearing load applied must be sufficient to compress the elastic bumper.

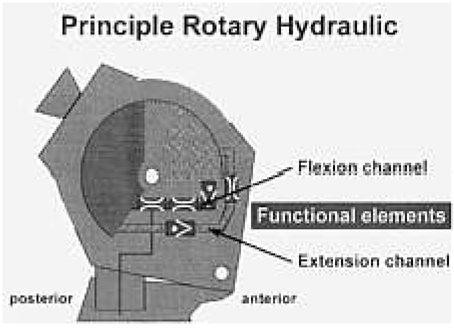


Figure 2. Functional diagram of the rotary hydraulic knee.

Stance Phase Functions

The position of the ground reaction force (GRF) with reference to the rotary and trigger axes determines the operating mode for this knee. Figure 3 illustrates the effect from these forces. When the GRF is anterior to both axes (F1), the joint moves into extension or remains fully extended against a fixed hyperextension stop.

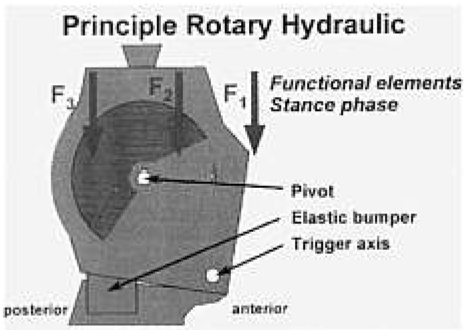


Figure 3. Ground reaction forces during walking determine the degree of stance control provided. F1 forces create a knee extension moment, and no stance control is needed. F2 forces cause the knee to pivot about the anterior trigger axis up to 4° thereby closing the stance control valve. F3 forces cause the knee to flex around the rotary axis; however, this motion is resisted by the hydraulic stance control feature.

As the GRF moves to a location between the two axes (F2), the Rotary Axis will remain extended fully while the joint will begin to flex about the trigger axis until there is equilibrium between the flexion moment caused by F2 and the counterbalancing moment from the compressible elastic bumper. This primary stance phase knee flexion is approximately four degrees.

Once the GRF moves posteriorly to both axes (F3), as long as the elastic bumper is compressed sufficiently the hydraulic stance control function will be activated. The opening of the stance flexion valve is automatically reduced and the resistance to the flow of fluid during further knee flexion is increased approximately tenfold. The knee will continue to flex under weightbearing load by rotating about the rotary axis so long as the GRF remains posterior to both axes.

Two adjustments are available to optimize the hydraulic stance flexion functioning of the 3R80 knee (Figure 4) . The lower control wheel varies the amount of hydraulic resistance to stance flexion, which may be described as the "stance control yielding rate." This is individualized according to the amputee's body weight and ability to control the knee with the residual musculature.

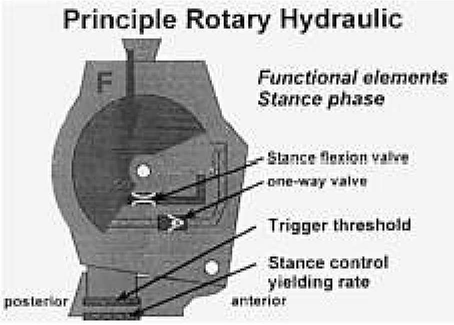


Figure 4. Stance control is adjusted with two thumb wheels. The upper adjustment wheel determines the amount of weight-bearing necessary to close the stance flexion valve. The lower adjustment wheel restricts the flow of fluid in flexion and sets the stance flexion yielding rate.

The upper control wheel varies the pre-compression of the elastic bumper, which then determines the amount of weightbearing loading required to close the stance flexion valve. This adjustment is optimized for the amputee's body weight and stride characteristics. When properly adjusted, the 3R80 will offer automatic stance control throughout the weight acceptance phase of gait, yet will also automatically disengage during preswing as the amputee's weightbearing gradually shifts to the leading (contralateral) limb. In contrast to earlier piston-based SNS units, the 3R80 user cannot inadvertently disengage stance control by stepping on the forefoot or accidentally generating a knee hyperextension moment. The 3R80 hydraulic stance stability is triggered consistently and predictably by the weightbearing load applied to the prosthesis.

Swing Phase Functions

For easy initiation of the knee flexion required for swing phase, it is critical that the stance control valve be fully opened (Figure 5) . When the GRF moves anteriorly to the trigger axis, the elastic bumper returns to its original height and the stance control valve automatically opens. Swing phase hydraulic resistances to both knee flexion and extension are independently adjustable via two separate valves. The 3R80 is designed to smoothly increase the resistance to knee extension as it nears 180 degrees; a hyperextension stop is also integrated into the mechanism.

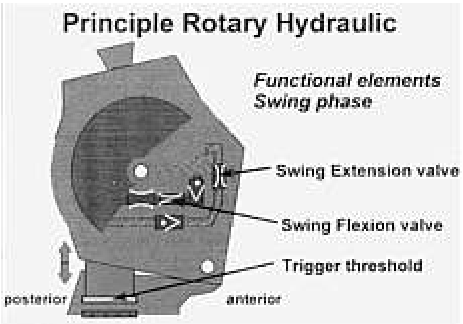


Figure 5. As soon as the GRF moves through or in front of the anterior trigger axis, the elastic bumper pushes the stance flexion valve open and disengages stance control automatically.

Gait Analysis

Methods and Test Subjects

The walking parameters of seven transfemoral amputees using the new 3R80 prosthetic knee joint were investigated using a previously published protocol.[4-7] In summary, all test subjects were successful long-term prosthetic users. A new prosthesis incorporating the 3R80 knee was designed, fitted, and aligned by a single experienced prosthetist (H.W. Scherer) in accordance with the manufacturer's recommendations. Data gathering occurred only when both the prosthetist and the amputee agreed that the prosthesis was optimized and the gait pattern appeared to be stable. This generally required several months of continuous use of the 3R80 prior to gait analysis.

Kinetic and kinematic data was gathered in an established gait laboratory using the PRIMAS system (Delft Motion Analysis, the Netherlands) which included dual Kistler force plates (Kistler AG Winterthur, Switzerland). External joint moments were calculated as the product of the perpendicular distance between the joint and the force line from the GRF multiplied by the magnitude of the GRF.

All subjects in this investigation were amputees secondary to traumatic injury, and none had any other major medical or physical deficits that might confound the results. The age at testing ranged from 28 to 59 years old with an average of 14 years of prosthetic use prior to receiving the 3R80. Details of the subjects' physical data are summarized in Table A .

SUBJECT	1	2	3	4	5	6	7
Socket	Ischial Containment	Ischial Containment	Ischial Containment	Ischial Containment	Ischial Containment	Ischial Containment	Ischial Containment
Suspension	Suction	Suction	Suction	Suction	Suction	Suction	Suction
Knee	3R80	3R80	3R80	3R80	3R80	3R80	3R80
Foot	1D25	1D10	1D10	1D10	IA30	1D25	IA30
Shoe	Jogging	Jogging	Jogging	Street	Jogging	Jogging	Street

Table A. Amputee data.

All test subjects were independent community ambulators capable of walking at least 5 kilometers daily. None used any walking aids, such as crutches, canes, or walkers. No subjects reported any significant pain that might interfere with walking although the majority admitted to occasional brief episodes of phantom pain. All subjects wore their customary type of shoe and prosthetic ankle-foot mechanism, in addition to the 3R80 knee. All were comfortably fitted with ischial containment sockets and suction suspension. Table B summarizes the prosthetic data.

SUBJECT	1	2	3	4	5	6	7
Age (years)	35	59	40	28	37	35	30
Mass (Kg)	76	63	96	75	91	64	71
Height (cm)	178	168	172	170	186	176	177
Cause	Trauma	Trauma	Trauma	Trauma	Trauma	Trauma	Trauma
Duration (yrs)	17	27	11	9	15	14	7
RL Length	Long	Long	Long	Medium	Medium	Long	Short
Side	Right	Left	Left	Left	Right	Right	Left
Walking Ability	>5km	>5km	>5km	>5km	>5km	>5km	>5km
Uses cane?	No	No	No	No	No	No	No

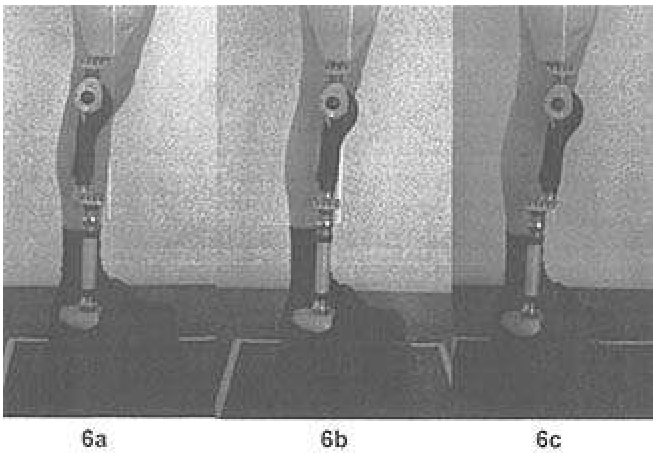
Table B. Prosthetic data.

The gait analysis portion of this investigation was designed to answer the following questions:

1. What are the characteristics of walking with this rotary hydraulic stance and swing control knee for this population?
2. What individual variations in the gait parameters are present?
3. What is the effect of sagittal plane linear alignment changes with this prosthetic knee?
4. What is the effect of varying prosthetic feet on the gait parameters with this knee?

In the first phase of the investigation, the amputees walked at a comfortable self-selected speed using the optimized prosthesis they had been successfully wearing full-time for many months. A minimum of nine gait trials were measured for all phases of the investigation and the mean values of these data are reported in this paper.

In phase two, all parameters were held constant except the linear position of the knee in the sagittal plane. After gathering data with the knee in the optimal position, the socket-to-foot relationship was held constant, but the knee was translated in 1 cm increments. Trials were conducted with the knee 2 cm and 1 cm posterior to the optimal alignment, optimally aligned, and 1 cm and 2 cm anterior to the optimal position. Figure 6 shows the special modified 4R103 slide adapter used above and below the knee mechanism to achieve isolated changes in the knee position.



6a 6b 6c

Figure 6. Specially modified slide adapters were used above and below the 3R80 knee, allowing the overall socket-

to-ankle/foot alignment to be held constant as the knee was shifted linearly. 6a shows knee in its most stable alignment (p-2cm). 6b shows the alignment considered subjectively optimal (p) as determined by the prosthetists during dynamic alignment trials. 6c is the most unstable position tested (p+2cm).

In the third phase, the effect of various prosthetic feet on walking with this knee was investigated. The optimal alignment using the amputee's preferred ankle-foot device was measured using the Otto Bock LASAR-Posture apparatus, which projects a vertical laser line from the center of pressure of the prosthetic limb during standing.[8] Six different prosthetic feet were then randomly interchanged without changing the linear alignment from that of the amputee's preferred foot. The flexion angle at the ankle was readjusted, as necessary, so that the center of pressure was identical to that measured with the preferred foot. This protocol was selected to eliminate the artifacts that might be created when a foot with a stiff keel was replaced with one having a more flexible forefoot, which might otherwise disturb the overall "balance" of the prosthesis during walking trials. The six feet tested with each amputee are summarized in Table C .

Generic Type	Name	Manufacturer
Elastic Keel	1D10 Dynamic	Otto Bock
Multiaxial Elastic Keel	1A30 Greissinger Plus	Otto Bock
Multiaxial Dynamic Response	1D25 Dynamic Plus	Otto Bock
Single Axis	1H38 Single Axis	Otto Bock
Multiaxial	Multiflex	Blatchford
Dynamic Response	Flex-Walk II	Flex-Foot

Table C. Prosthetic feet tested.

Results

All amputees completed all trials without difficulty, walking at a self-selected speed between 1.1 m/s and 1.5 m/s with an average of 1.3 m/s. This data is in agreement with previously published walking speeds for traumatic transfemoral amputees.

Walking with optimized definitive prosthesis with 3R80 knee. The knee angle plots of both the amputated and contralateral sides showed clear inter-individual variations throughout the gait cycle (Figure 7) . The maximum flexion angle of the prosthesis during midswing is usually greater than the maximum midswing value for the contralateral leg. Examination of the knee extension angles near the end of swing phase confirms the smooth terminal damping designed into the 3R80 and shows that the prosthetic knee is always in full extension just prior to the next gait cycle.

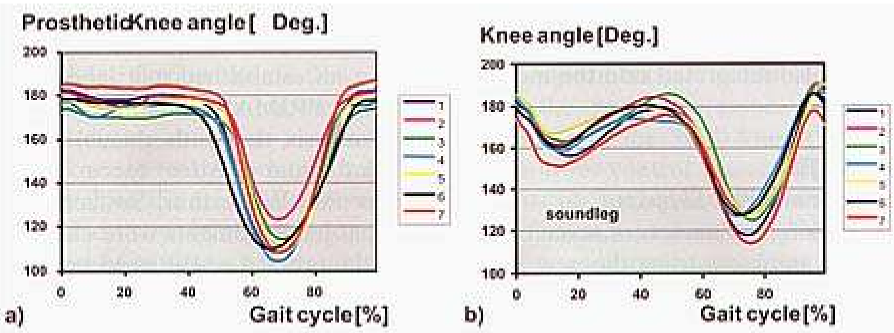


Figure 7. Plot of knee flexion angle for both prosthetic (a) and sound side (b) for all subjects. The amount of stance phase knee flexion near 20% of the gait cycle is reduced on the prosthetic side while the degree of maximum swing flexion is usually slightly higher than on the contralateral side.

Note the apparent variation in the initial and terminal knee angles from the expected 180 degrees. In fact, the prosthetic knee was as fully extended as is mechanically possible. Slight differences in the sagittal plane alignment of the prosthetic knee, based on the amputee's clinical needs, resulted in a knee marker that was not always located directly on the hip-ankle marker line. This created the appearance of a knee that is flexed more or less than 180 degrees at these times in the gait cycle.

Figure 8 takes a closer look at prosthetic knee motion data during the first half of the gait cycle. The knee angle at heel contact has been normalized to 180° for this comparison. Note that the maximum amount of stance phase knee flexion for well aligned prostheses measured between 2° and 8° for this population.

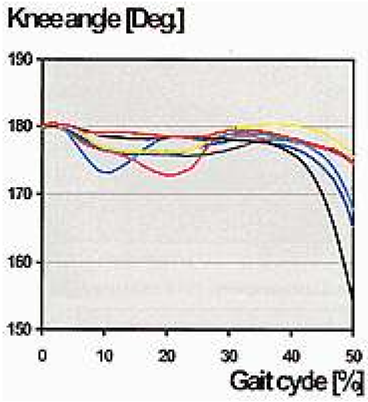


Figure 8. Knee angle plot for all seven subjects shows the amount and timing of stance phase knee flexion varied. Full knee extension equals 0°.

Figure 9 provides an overview of the external joint moments while walking with the definitive prosthesis, optimally aligned with the 3R80 knee. Moments for the prosthetic limb and the contralateral knee are of interest. As would be expected, the average external moment at the prosthetic ankle during the early portion of the gait cycle is toward plantarflexion (Figure 9c) . It gradually changes to a dorsiflexion moment for the rest of stance phase.

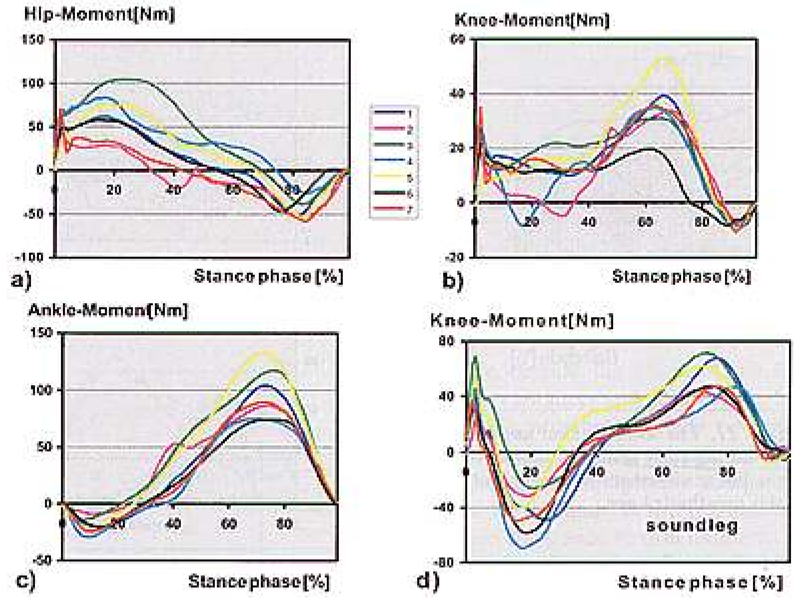


Figure 9. External joint moments at the subjects' self-selected walking speeds with an optimally aligned 3R80 knee.

The prosthetic knee moments are more variable (Figure 9b) . Two amputees (subjects 2 and 4) walk confidently despite significant flexion moments at the knee during weight acceptance; the peak knee flexion occurs at different times in the gait cycle, however. The knee extension moments during terminal stance also vary among the individual subjects. For example, the slope of the knee moment between 40% and 60% of stance phase is very steep for subject 5, indicating a rapid increase in the extension moment. In contrast, the slope of the knee moment during the same period for subject 6 is much more horizontal, indicating a more gradual increase in the extension moment as well as a much lower maximum knee extension moment.

The hip moments on the prosthetic side are also highly variable (Figure 9a) . The maximum hip extension moment for stance stability varies between 25 and 105 Nm, while the maximum hip flexion force necessary to initiate pre-swing flexion ranges from 27 to 59 Nm. One amputee (subject 2) demonstrated an abrupt reversal in the hip moment at about 45% of stance phase; the curves of the other six subjects did not have this pattern.

Patterns for the contralateral knee moments are similar among the subjects but vary somewhat in both magnitude and timing (Figure 9d) . Presumably, this is due to the influence of active muscle control on the biological knee as well as the varying need to compensate for the amputation with the surviving leg. Movements of the prosthetic knee are much more predictable, despite the effect of the hydraulic resistances and adjustments, because the prosthetic device responds primarily to the external moments at the knee. For example, by comparing Figure 9b to Figure 8 , the relationship between the stance flexion moments of subjects 2 and 4 and the resulting stance flexion angle of the knee become clear. The approximate relationship between the timing and magnitude of the moment and the actual knee movement can also be discerned.

Linear sagittal alignment of the 3R80 prosthetic knee joint. Phase one earlier demonstrated significant inter-individual variations in many gait parameters. As a result, simple mean values are not of great interest when investigating discrete changes to the prosthesis. For this reason, the phase two investigation of linear knee placement reports the results of various knee alignments for a single amputee subject.

For this average-size adult amputee, shifting the knee 1 cm changes the knee angle as measured by the body markers approximately 2.5 degrees. As reported earlier in this paper, the knee was offset posteriorly and anteriorly from the optimal position in 1 cm increments. Figure 10 demonstrates that the prosthetic knee angle curve is offset rather uniformly by these changes while the contralateral biological knee curve is largely unaffected. This suggests minimal compensation for prosthetic knee alignment changes in sound side knee motion.

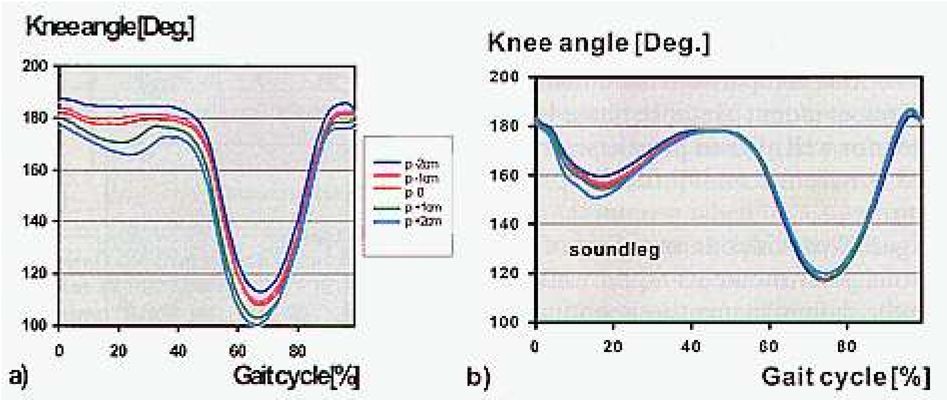


Figure 10. Prosthetic knee angle curves (a) vary throughout the gait cycle with differing sagittal plane alignment of the knee mechanism. In contrast, the sound side knee angle curves (b) vary only in early stance between 5% and 25% of the gait cycle.

Figure 11 offers a closer look at the stance phase of gait for this subject. With the optimal alignment, this amputee demonstrates a maximum stance flexion knee angle of 4°. Moving the knee 2 cm posteriorly decreases the stance flexion maximum to 2°; moving it 2 cm anteriorly increases the stance flexion maximum to 10°. Alignment changes of 1 cm give intermediate results. Thus, it appears that the stance flexion function of the 3R80 knee is directly related to linear alignment changes in the sagittal plane.

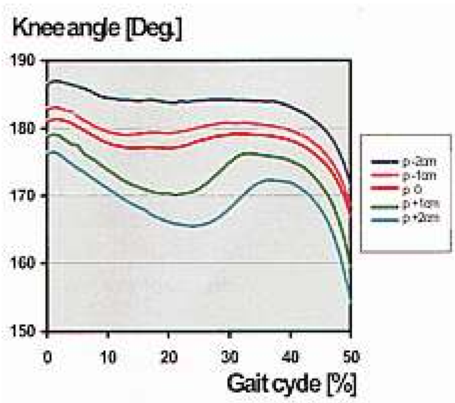


Figure 11. The knee flexion angle of the 3R80 during early stance is directly related to the linear alignment in the sagittal plane of this prosthetic knee.

Figure 12 shows the external moments resulting from these alignment changes. The effect of linear sagittal alignment changes on the prosthetic knee moments (Figure 12b) is most striking, demonstrating clearly that as the prosthetic knee mechanism is moved in a posterior direction, the knee extension moments increase proportionally: the more posterior the knee, the more stable the prosthesis. On the other hand, anterior movement of the knee creates proportional instability as the knee extension moment decreases or even changes to a small flexion moment. Thus, it appears that the magnitude and timing of the external knee moments cause the stance flexion function of the knee to vary.

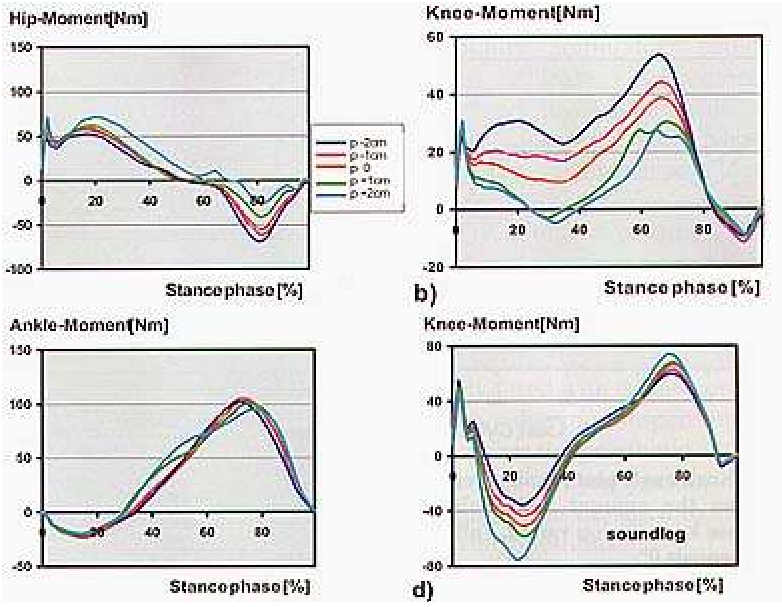


Figure 12. The external moments vary significantly as the sagittal alignment of the 3R80 is changed. In 12a, during weight acceptance (at approximately 20% of stance phase), the hip extension moment increases significantly with each anterior displacement of the knee. 12b shows the prosthetic knee extension moment is decreased as the alignment is varied from 2cm posterior to optimal up to 2cm anterior to optimal. 12c indicates that when the stance flexion angle s larger, the forefoot of the prosthetic foot contacts the ground more rapidly and leads to an earlier increase in the ankle dorsiflexion moment. 12d demonstrates that the external moments of the biological knee change primarily in early stance.

The hip moment the amputee must generate is also related to prosthetic knee stability. Figure 12a shows that during weight acceptance at approximately 20% of stance phase, the hip extension moment increases significantly when the knee is moved 2 cm anteriorly and slightly when it is moved ahead 1 cm. However, the hip flexion moment required during preswing (at approximately 80% of stance phase) is reduced when the knee is shifted anteriorly, making it easier for the amputee to initiate flexion of the prosthetic device. In contrast, when the prosthetic knee is translated posteriorly to the optimal position, the hip extension moment in early stance is largely unchanged, but a greater hip flexion moment is required during preswing. It is interesting to observe that, for this population, the hip moments change significantly with each incremental change in prosthetic knee position even though the knee moments are similar for both 1 cm and 2 cm of anterior translation.

Figure 12c documents that the prosthetic ankle moment is only affected by larger magnitudes of stance phase knee flexion. When the stance flexion angle is larger, the forefoot of the prosthetic foot contacts the ground more rapidly. This leads to an earlier increase in the ankle dorsiflexion moment. Figure 12d illustrates the moments at the contralateral, biological knee. The biological knee moments change primarily in early stance, when the flexion moment increases as the prosthetic knee is moved in an anterior direction. In general, the contralateral hip and ankle moments did not vary significantly so this data is not presented here.

Effect of various prosthetic ankle-foot mechanisms. During the third phase of this study, as noted earlier, several commonly used prosthetic feet were randomly interchanged in the definitive prosthesis, and the gait parameters were measured. Prior to data gathering, the amputees were allowed to walk as long as they felt was necessary to acclimate to the new foot.

The choice of prosthetic ankle-foot device significantly influences subjective walking comfort for the amputee as well as the external moments about the prosthetic knee, depicted in Figure 13b . The moments about the ankle joints (Figure 13c) are also dependent upon the foot type, as has been noted in previous publications.[4] The hip moments are generally unaffected by the foot variation with the exception that the Flex-Walk II resulted in a higher hip flexion moment during preswing at approximately 80% of stance phase (Figure 13a) . Sound side knee moments (Figure 13d) are also very similar for all tested mechanisms with the exception of the Flex-Walk II.

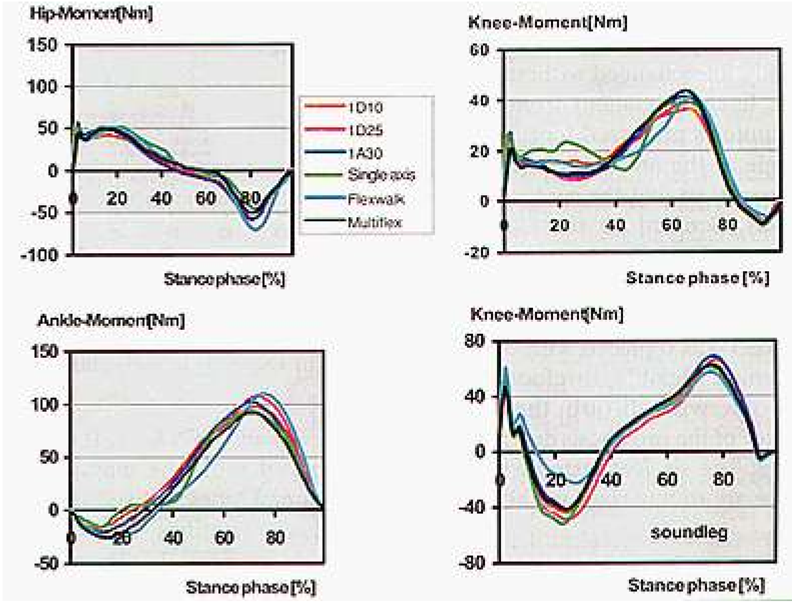


Figure 13. This figure illustrates the effect of differing ankle-foot mechanisms on the joint moments during stance phase. Hip moments (a) are largely unaffected with the exception that the use of the Flex-Walk II results in a higher hip flexion moment at about 80% of stance phase during preswing. 13b shows the prosthetic knee moments in the first half of stance phase are significantly affected by the ankle-foot device. The prosthetic ankle moments (c) also vary according to type of foot mechanism used, and sound side knee moments (d) are very similar for all tested mechanisms with the exception of the Flex-Walk II.

Table D summarizes the subjective preference of each amputee subject at the conclusion of the three studies for the specific feet that were tested. With 1 being the most preferred foot and 6 being the least desired foot, it is interesting to note that the mean ranking for all feet was near the mid-range of 3-4, with a variance from 2.7-4.7 overall. This reflects the common clinical observation that many prosthetic ankle-foot mechanisms are available and it is often difficult to predict amputee preference with any accuracy. Yet, the ease with which each subject ranked the options suggests that differences are significant to the individual and lead to the selection of one foot over another.

Subject	1	2	3	4	5	6	7	Mean Rank
1D25	1	5	2	2	3	1	5	2.7
1A30	2	4	3	4	1	4	2	2.9
Flex-Walk II	4	3	6	1	5	2	1	3.1
1D10	3	6	1	5	2	3	6	3.7
Multiflex	5	1	5	3	4	5	4	3.9
1H38	6	2	4	6	6	6	3	4.7

Table D. Amputee foot preference. (1 = First Choice, 2 = Second Choice...,6 = Sixth Choice.)

This data suggests some trends in these transfemoral amputees' choice of ankle-foot mechanisms. In general, there is a preference for the highest energy return option available-but only so long as there is no need for compensation with the sound side limb in terms of joint moments. Figure 14 compares the data from subjects 7 and 3. The graphs for patient 7 show that the prosthetic ankle joint moment with the Flex-Walk II is clearly higher at 80% of stance phase than with all other alternatives, yet the contralateral knee moments are unchanged. In this circumstance, the carbon fiber dynamic response foot will be preferred, as Table D verifies.

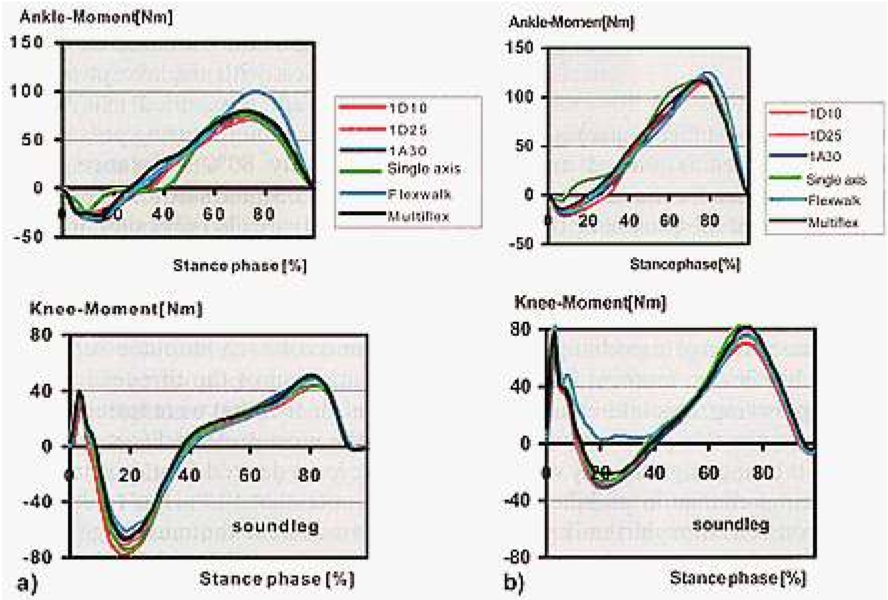


Figure 14. This figure compares data from subject 7 (left column = a) and subject 3 (right column = b). For subject 7, the prosthetic ankle joint moment with the Flex-Walk II is clearly higher at 80% of stance phase than with other tested alternatives, yet the contralateral knee moments are unchanged. In contrast, subject 3 demonstrates no major changes in ankle moments when using the Flex-Walk II while the contralateral knee moments between 10% and 40% of stance phase suggest the biological knee may be compensating for the characteristics of this foot device.

In contrast, the data for subject 3 in Figure 14 shows no major change in prosthetic ankle moments with the Flex-Walk II. Even more important, the contralateral knee moments between 10% and 40% of stance phase suggest that the biological knee is compensating for the characteristics of the Flex-Walk II foot. As seen in Table D , this subject rated the Flex-Walk II as the least desirable alternative for his prosthesis. Subjects who ranked the Flex-Walk II between these two extremes demonstrated intermediate gait parameters as well.

One additional biomechanical criterion for the choice of prosthetic ankle-foot devices appears to be the amputee's ability to generate a hip extension moment in early stance to stabilize the prosthesis. Subjects with a relatively low hip extension moment during weight acceptance prefer feet that produce minimal knee flexion moments during this phase of the gait cycle. Referring to Figure 9 for data about subject 2, it is clear that he demonstrated the lowest hip extension moment of all subjects tested. Figure 15 compares the moments and angles for the 3R80 prosthetic knee using this subject's first and last choice of prosthetic feet. The decreased knee moments generated by the flexibility of the multiaxial Multiflex foot creates a smoother knee angle plot that allows for a more controlled walking pattern.

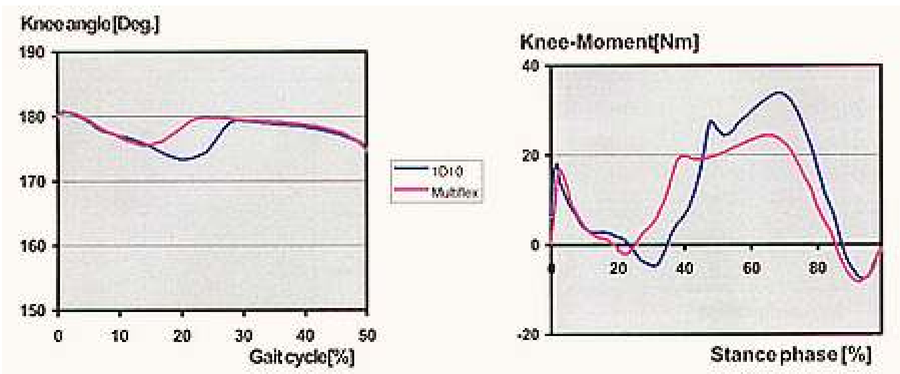


Figure 15. Comparison of the moments and angles for the 3R80 knee when subject 2 is using his most and least preferred ankle-foot mechanisms.

We postulate that intermediate ankle-foot choices may also be influenced by these biomechanical factors. If the need for contralateral knee compensation or the lack of any ankle moment advantage rules out the dynamic response feet, then the amputee's ability to generate a hip extension moment during the weight acceptance phase of gait appears to become the critical factor. For the feet tested in this study, those transfemoral amputees with a strong hip extension moment (but who did not want a dynamic response foot) chose the flexible keel 1D10 Dynamic foot. Those with a moderate hip extension moment selected feet with a multiaxial ankle: either the 1D25 Dynamic Plus or the 1A30 Greissinger Plus. As previously noted, those with a low hip extension moment seemed to prefer feet that generate a minimal flexion moment, such as the Multiflex or the single axis 1H38 ankle-foot.

This logic appears to confirm the subjective choices of the amputees in this study. Obviously, further investigations over much larger populations are necessary to determine if these trends are significant.

Summary of the biomechanics of walking with the 3R80 prosthetic knee joint. During swing phase, the gait pattern of transfemoral amputees using the 3R80 prosthetic knee is similar to that of other hydraulic swing phase control devices studied previously in the literature.[9,10] The only exception is that the integrated pre-extension damping generates a smoother knee angle curve in terminal swing as the prosthesis reaches full extension Figure 7a .

Stance phase functioning is determined primarily by the changes in ground reaction forces (Figure 16) . At initial contact, the GRF is well anterior to the knee. During weight acceptance, the GRF moves quickly to a position between the two joint axes and mechanical stance flexion of approximately 4° occurs. On level ground, the GRF approaches the rotary axis but does not always pass further posterior. During midstance, the GRF again moves more anteriorly until terminal swing. During preswing, the 3R80 knee is easily flexed despite partial weightbearing whenever the GRF passes at or posterior to the rotary axis.

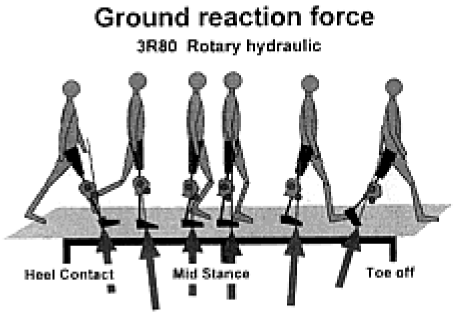


Figure 16. Overview of the ground reaction force throughout the gait cycle and its position with reference to the anterior trigger axis and rotary axis. Stance stability is determined by the amount of weightbearing as well as by the biomechanics of amputee gait.

Discussion

The Mauch and CaTech hydraulic cylinders provide both stance and swing phase control in one unit. With these devices, switching from one mode to the other occurs whenever a hyperextension moment is applied to the extended knee for more than one tenth of one second. Although this moment usually occurs in late stance phase when walking on level ground, it can also occur inadvertently at undesirable times. For example, when descending stairs the novice amputee often forcefully hyperextends the hip thereby eliminating stance phase control. On irregular surfaces, sudden loading of the forefoot while the knee is extended can also disengage stance control by generating a similar hyperextension moment. Disengagement of the stance control feature by a hyperextension moment occurs regardless of the amount of weightbearing applied to these earlier knee designs.

This article has summarized the gait parameters of a group of traumatic transfemoral amputees using a new rotary hydraulic knee where the weightbearing load determines which mode is functioning. Unlike earlier hyperextension-triggered units, the 3R80 is not susceptible to inadvertent loss of stance stability. In addition, the amount of weightbearing load required as well as the flexion yielding rate can be individualized to complement the amputee's physical capabilities.

The moment created by the hydraulic stance resistance of the 3R80 knee is constant throughout the full range of 135° of flexion (as long as the loading on the knee is constant). In contrast, the yielding resistance of earlier cylinder-based designs gradually decreases as knee flexion progresses due to the lever arms between the piston rod attachment and the knee center of rotation.

Gait analysis of a group of seven amputees using this new prosthetic knee design demonstrates that most preferred an alignment that offered mechanical stance flexion on level surfaces while reserving the fluid-controlled stance flexion range for descents, such as on ramps or stairs. Amputees who walk with a very slow cadence may find the unlimited range of stance flexion in the 3R80 disconcerting. Individuals who walk slowly will often prefer a friction brake knee, such as the 3R49, or a polycentric stance flexion knee, such as the 3R60.[6,7]

The large range of hip moments measured in this group appears to be the result of individual physical variations. Interchanging various prosthetic feet or changing the sagittal alignment of the 3R80 had little effect on the hip extension moment of a particular amputee unless the knee was more than 1 cm anterior to its optimal position and was therefore significantly unstable. The hip flexion moments, in contrast, varied directly with the sagittal alignment: the more stable the alignment, the greater the hip flexion moment required by the amputee to initiate knee flexion in preparation for swing phase.

Although the 3R80 knee was suitable for use with a variety of prosthetic ankle-foot mechanisms, each amputee had clear (and differing) preferences for specific foot devices. Previous researchers have shown that the choice of prosthetic feet can alter the biomechanics of both the prosthetic and contralateral limbs.[4,11,12] This study identified a trend toward rejection of a carbon fiber, dynamic response foot when there was no ankle moment benefit or when it was necessary to compensate with the contralateral knee. In this situation, it appears that each amputee then chooses a particular prosthetic foot to complement their individual ability to generate a hip extension moment on the prosthetic side. The lower the transfemoral amputee's ability to generate a hip extension moment in early stance, the more they seem to prefer an ankle-foot device that generates a minimal prosthetic knee flexion moment.

The results of this study, although limited by the small number of subjects tested, suggest that the well-recognized clinical variability in amputee preference for prosthetic components and specific alignments may have a fundamental biomechanical basis. The investigation of various alignments and component combinations on objective gait parameters revealed that knee alignment stability has a significant, primary, and predictable effect on the amputee hip flexion moment required during the transition from stance to swing phase. Other variations in moments or motion appear to be attributable to physical differences or the direct results of the biomechanics of the prosthetic limb.

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